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Enginyeria en Tecnologies Industrials

Gait analysis of paediatric patients with hemiparesis

MEMÒRIA

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Abstract

The main objective of this Final Project for the Bachelor Degree in Industrial Technology Engineering is to study how the feedback device Walking o'Clock modifies gait pattern of paediatric patients with hemiparesis. The project has been developed in collaboration with personnel of Sant Joan de Déu Hospital (HSJD), that selected the three patients involved in the study. The gait of these patients was captured in the UPC Biomechanics Laboratory, and the kinematic analysis was performed using OpenSim, a free software tool developed by Stanford University that is widely used by the scientific community.

Walking o'Clock, by Draco Systems, is an electronic device with an inertial measurement unit (IMU). It was used to measure thigh orientation in the study, aided by the engineer who created this product. By measuring this orientation, the physiotherapist would choose what kind of feedback the patient should be put under to.

The patients' movement was analysed under three different situations: natural gait, gait using the device (with the feedback chosen) and gait after using the device (after feedback). Four angular coordinates in the sagittal plane (hip flexion, pelvic tilt, knee flexion and ankle dorsiflexion) were analysed and compared.

From the results, it was shown that the device modifies the gait pattern. However, depending on the patient and the feedback, the walking kinematics was modified in different ways. In some aspects, an improvement was found for the selected paediatric patients.

This report describes all the processes involved in the analysis, as well as the methodology used. To obtain the motion data, the human body has been modelled as a multibody system with rigid bodies and ideal joints with different degrees of freedom. The process to export the kinematics data using OpenSim is explained in detail. From the position of each body, inverse kinematics determines the configuration (position and orientation) of the multibody system along time.

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1. Glossary

There are some terms that are used with their acronyms in the project. Note that they will be defined later on the project. The list is presented here:

- GC: Gait Cycle
- IC: Initial Contact
- IK: Inverse kinematics
- DOF: Degree of freedom
- N: Natural
- F: Feedback
- F1: Feedback 1
- F2: Feedback 2
- A: After Feedback
- ROM: Range of Motion
- RMSE: Root Mean Square Error

2. Introduction

This Final Project for the Bachelor's degree in Industrial Technology Engineering is entitled "Gait analysis of paediatric patients with hemiparesis", and is part of the line work of the Biomechanical Engineering Group (BIOMECH) of the Department of Mechanical Engineering at the Barcelona School of Industrial Engineering, at Universitat Politècnica de Catalunya (ETSEIB, UPC).

This project has been developed in collaboration with Sant Joan de Déu Hospital (HSJD), specifically with the Research Team, that has provided with the patients involved in the study, and with Draco Systems, the engineering company that has designed the device used in the study. The results and conclusions of this project will be used by the HSJD for further investigations.

2.1. Origin of the project and motivation

First of all, the author of this project wanted to do a project related to biomechanics and, if possible, with some kind of contribution to a research area, since she will be coursing the Master in Industrial Engineering with the speciality in Biomedicine.

Second, the Research Centre of HSJD does not have a biomechanics laboratory on their facilities. Therefore, months ago, they approached the Biomechanical Engineering Group of the UPC to see if they could have any kind of collaboration in the future.

At the same time, an engineering company, Draco Systems, had designed a prototype of a device (named Walking o'Clock), under the demands of the HSJD, which would assist and modify the gait pattern of paediatric patients with a walking disability. However, it had not been quantitatively assessed yet.

After a couple of meetings between the HSJD team and the BIOMECH team, the idea of the project was decided. It would be a pilot study on how this device modified the gait pattern of paediatric patients with hemiparesis, who would be selected by the clinical staff of HSJD. Using the equipment in the Biomechanics Laboratory in the university (ETSEIB), captures of gait would be taken. Then, the measurements would be analysed with the software OpenSim and the results would be compared to those of a healthy gait.

2.2. Introduction to Biomechanics

Biomechanics is the study of the structure and function of biological systems such as humans, animals, plants, organs, fungi, and cells by means of the methods of mechanics [1].

In this project, the kinematics of human gait will be analysed. Due to the complexity of human

body, formed by multiple systems and tissues, some simplifications are made to analyse movement. The human skeleton is in charge of the movement from joints between bones, which is possible thanks to the muscles, which coordinate their action from the orders generated by the central nervous system.

Musculoskeletal models are effective for visualizing human movement, analysing the functional capacity of muscles, and designing improved surgical procedures [2]. They model the human body as a system formed by several solids or rigid segments, joined together through joints that allow different degrees of freedom. This includes the study of the kinematic and dynamic motion of human body in the field of rigid body mechanics.

The goal of this project is to carry out a pilot study on how a device modifies the gait pattern. This device, named Walking o'Clock, is still a prototype, whose operation will be explained in detail further in this work. Three patients, selected by the clinical staff of Hospital Sant Joan de Déu, will be analysed independently. Simultaneously, their gait patterns will be compared to the standard one [3]. Their gait pattern will be analysed from motion captures taken in the Biomechanics Laboratory of the Technical University of Catalonia, UPC, in ETSEIB. After processing the data from the captures, the kinematic analysis is done through the free software *OpenSim* [2], developed by the *National Center for Simulation in Rehabilitation Research* of the Stanford University (California, USA).

Child patients with hemiparesis attend physiotherapy, as often as their pathology demands. However, it is never as much as they need. This means that, after assisting therapy, the aspects that they have worked improve for a few days, but tend to go back to their original state if therapy is not repeated soon enough. The impossibility of having a training with a physiotherapist every day, as they would need, complicates and slows the healing process. Therefore, if patients could have something that complemented the therapy, the process would be improved. This is where the device makes sense. Since it would be affordable, it would allow patients to have a daily assistant. They could use it every day, whenever and wherever they wanted. Of course, the use of the device would be controlled and supervised by their physiotherapist.

2.3. Objectives of the project

The general goal has been defined, in accordance with the HSJD team and the UPC Biomechanics group, as the study of how a device modifies a gait pattern in three paediatric patients with hemiparesis. These general goal is developed into these specific objectives:

- Analysing the gait of three paediatric patients with hemiparesis in three different situations, using the UPC Biomechanics Laboratory, and investigating how the Walking O'Clock device modifies the gait pattern of each patient.
- Finding a model and a protocol to reproduce the motion as close to reality as possible.
- Obtaining kinematic variables of the gait using the OpenSim software.
- Comparing the results of the three different situations for each patient, and comparing them to the standard healthy gait to quantify the change achieved by using the device.
- Reporting the results obtained to the Research Team of HSJD.

2.4. Project Scope

The aim of this pilot study is to analyse how the device modifies the gait pattern of three children patients with hemiparesis. Anyhow, it is important to keep in mind the timing and circumstances of the study. First of all, each patient will have used the device for a short time, so they will not have been able to adapt to its functioning as well as they would have been in case of having used it for a longer time. Secondly, due to the fact that the device is a prototype, it has some flaws and it might not be completely reliable. Moreover, the physiotherapist involved with the project is still getting used to the device. However, the motion captures that will be taken will show how the gait pattern of each patient changes after using it, so this changes will be analysed with a more accurate and reliable method. Therefore, even if the device has some error, it will not affect the analysis of the consequences of using it.

Then, the main goal of this project is to show how the use of this device would modify the gait pattern of a patient and to see if the changes improve the gait. It is then a pilot study for what could be a deeper study with more patients and longer time.

3. Background

3.1. Biomechanics of human gait

3.1.1. Reference planes

The motion of limbs is described using three reference planes, which can be seen in Fig. 1. The three reference planes of the anatomical position. The sagittal plane is any plane which divides the part of the body into right and left portions; the median plane is the midline sagittal plane, which divides the whole body into right and left halves. The frontal plane divides a body part into front and back (anterior and posterior) portions. Finally, the transverse plane divides a body part into upper and lower portions [4]. In this project, only the motion patterns in the sagittal plane will be analysed.

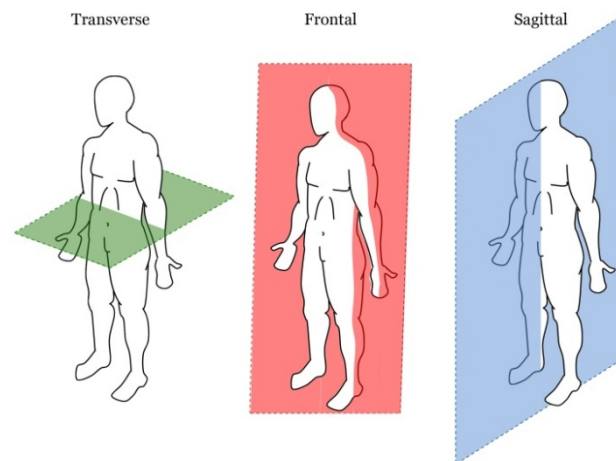


Fig. 1 The three reference planes of the anatomical position [5]

3.1.2. Gait cycle

Human gait is the result of the complex interaction of several subsystems, the neuromuscular, musculotendinous and osteoarticular, that generate together the necessary body dynamics for bipedal movement. Walking uses a repetitious sequence of limb motions to simultaneously move the body forward while also maintaining stance stability. As the body moves forward, one limb serves as a mobile source of support while the other limb advances itself to a new support site. For the transfer of body weight from one limb to the other, both feet are in contact with the ground. A single sequence of these functions by one limb is called a gait cycle (GC).

Each GC is divided into the stance period, during which the foot is on the ground, which begins with an initial contact (IC), and the swing period, when the foot is in the air for limb

advancement, which begins as the toe is lifted from the floor (toe off). The GC begins with initial double stance (bilateral foot contact with the floor), followed by the single limb support as the opposite foot is lifted for swing (single stance, SLS). The duration of such phase is the best index of the limb's support capability, with longer relative durations reflecting greater stability. The third subdivision is the terminal double limb stance, which begins with floor contact by the other foot and continues until the original stance limb is lifted for swing.

The generic normal distribution of the floor contact periods approximates 60% for stance and 40% for swing, which varies with the person's walking velocity [4]. It shows an inverse relationship to walking speed, so the change in stance and swing times becomes progressively greater as the speed slows. An opposite relationship can be found among the subdivisions of stance, as walking faster proportionally lengthens single stance and shortens the two double stance intervals. The subdivisions in gait are shown in Fig. 2.

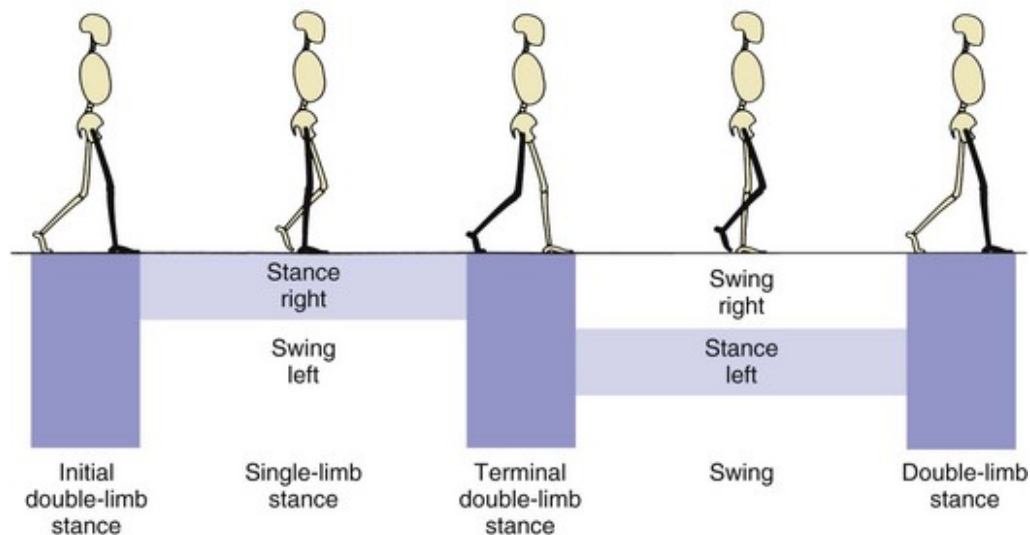


Fig. 2 . Subdivisions of gait and their relationships to the pattern of bilateral floor contact (Adapted from [6])

Since one action flows into the next, there is no specific starting or ending point. At the Biomechanics Laboratory in the university, measures of complete GC for each limb can be taken. Taking into account that normal people initiate floor contact with their heel, as the right feet is the first one to step on the pressure plate, its captures cycle will be from heel strike to heel strike (HE-HE). On the other hand, the captured cycle for the left limb will start a few instants later, going from toe off to toe off (TO-TO).

The GC has also been identified by the descriptive term stride, which is the period from initial contact to initial contact of the same limb, which comprises two steps. The step is the distance between initial swing and initial contact of the same limb. The relationship between step and stride is seen in Fig. 3.

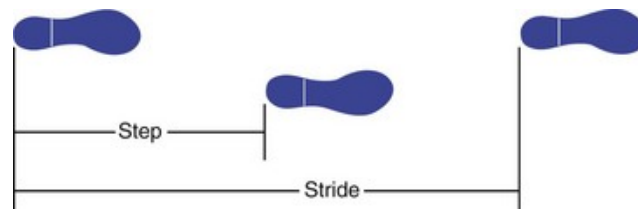


Fig. 3 Step versus stride (Adapted from [6])

Phases of Gait

Analysis of a person's walking pattern by phases more directly identifies the functional significance of the different motions occurring at the individual joints [4]. This is important for interpreting the functional effects of disability. The limb must accomplish 3 basic tasks: weight acceptance (WA), single limb support (SLS) and swing limb advancement (SLA).

As seen in Fig. 4, the periods show the basic division of GC by foot contact. Each phase is determined by limb postures. The tasks show the grouping of the phases by the functions to which they contribute [4].

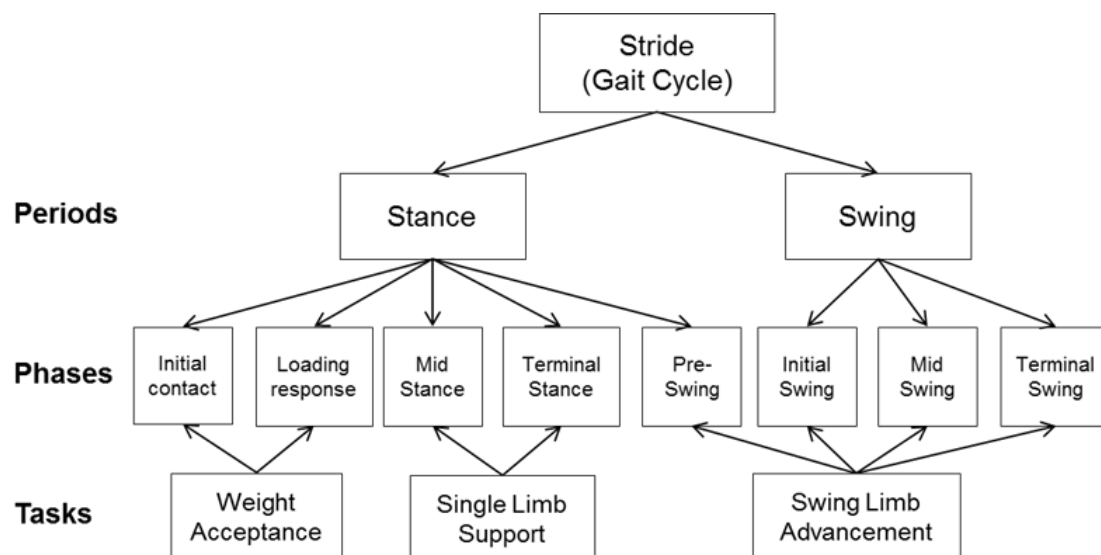


Fig. 4 Functional division of the GC

Weight acceptance

Being the most challenging task in the GC, 3 functional demands must be satisfied: shock absorption, initial limb stability and the preservation of progression, all of which result in transferring the body weight onto a limb that has just finished swinging forward and has an unstable alignment. The IC and loading response phases are involved.

Single limb support (SLS)

The SLS interval for the stance limb begins when the contralateral foot takes off for swinging

and continues until the opposite foot again contacts the ground. In this case, the one limb has the total responsibility of supporting the body weight in both the sagittal and coronal planes while progression continues. Mid stance and terminal stance are involved in SLS.

Swing limb advancement

Preparatory posturing begins in stance in order to meet the high demands of advancing the limb. Then, the limb swings through 3 postures as it lifts itself, advances to complete the stride length and prepares for the next stance interval. Four gait phases are involved: pre-swing (end of stance), initial swing, mid swing and terminal swing.

Determinants of gait (motion patterns)

Fig. 5 shows sagittal plane joint angles (in degrees) during a single gait cycle of pelvic tilt, right hip (flexion positive), knee (flexion positive) and ankle (dorsiflexion positive).

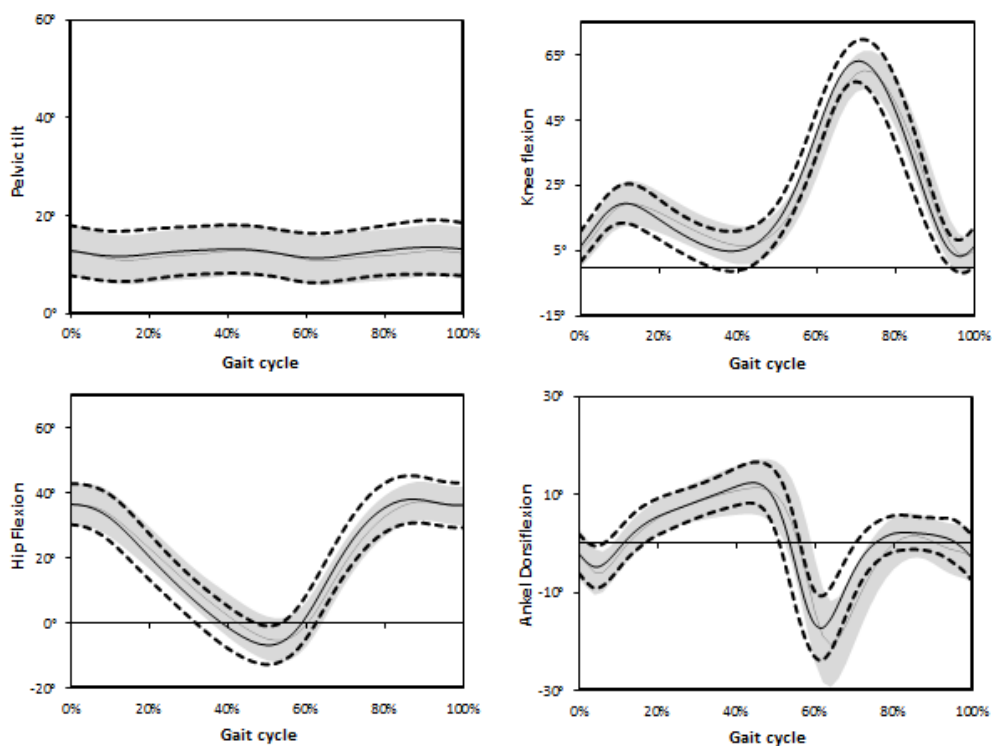


Fig. 5 Sagittal plane joint angles during a single gait cycle of pelvic tilt and right hip, knee and ankle [7].

3.1.3. Gait analysis

Gait analysis is used for two very different purposes: to aid directly in the treatment of individual patients and to improve our understanding of gait, through research. No single method of analysis is suitable for such a wide range of uses and a number of different methodologies have

been developed [4].

Basically, there are 5 measurement systems. Three of these focus on the specific events that constitute the act of walking. Motion analysis defines the magnitude and timing of individual joint movements. Dynamic electromyography identifies the period and relative intensity of muscle function. Force plate recordings display the functional demands being experienced during the weight-bearing period. Each system serves as a diagnostic technique for one facet of gait. The 2 remaining gait analysis techniques summarize the effects of the person's gait mechanics. One measures the patient's stride characteristics to determine overall walking capability, while efficiency is revealed by energy cost measurements. There are several choices of technique within each of these 5 basis measurements systems, which differ in cost, convenience and completeness of the data provided. As there is not a single optimal system, selections are based on the needs, staffing, and finances of the research situation [8].

3.2. Cerebral palsy

Cerebral palsy is defined as a group of permanent disorders in the development of movement and posture causing activity limitations that are attributed to non-progressive disturbances that occurred in the developing foetal or infant brain [3]. Along with the motor disorders, disturbances of sensation, perception, cognition, communication and behaviour can appear, as well as epilepsy or secondary musculoskeletal problems. Not only affects it the cortical brain, but all of the brain's functions as well. In the case of gait, it influences the musculoskeletal system function.

The characteristics of cerebral palsy were given by Gage [3] as follows:

1. Loss of selective muscle control
2. Dependence on primitive reflex patterns for ambulation
3. Abnormal muscle tone
4. Relative imbalance between muscle agonists and antagonists across joints
5. Deficient equilibrium reactions

It is important to keep in mind that the way cerebral palsy affects the joint rotation varies considerably between one patient and another.

3.2.1. Children with cerebral palsy

Spasticity refers to stiff or rigid muscles, an overactive response of a muscle to rapid stretching. It can also be called unusual stress or increased muscle tone. Reflexes are stronger or exaggerated. This condition can interfere with the activity of walking, movement or speech [9].

Even if children with cerebral palsy do not have severe spasticity or weakness, they usually

have problems with motor control, including difficulty in moving their limbs. Moreover, balance is a significant problem in most children. The aetiology of this abnormality is not clear, but it is probably secondary to poor connections between the cerebellum and most of the other higher centres of the brain, as well as connections to descending neurons to the spinal cord.

Paediatric hemiparesis

Paediatric spastic hemiparesis is a mild form of cerebral palsy in children, which affects only half of the body. It is a neurologic condition which complicates movement in that part of the body, but without reaching paralysis. Therefore, it is a minor degree of hemiplegia, which is total paralysis and affects only one out of thousand born children [10].

The most typical manifestation consists on abnormal postures and limb deformities due to weakness of some muscles and spasticity of others. In the upper limb this results in a difficulty in performing manual activities; the typical stance is with an elbow, wrist and fingers flexion. At the bottom, movement (gait) problems can be found, being the typical stance with flexed knee and ankle [11].

Hemiparetic gait patterns

In the case of hemiparesis, the usual gait pattern is when the patient walks slowly, resting their weight on the unaffected limb, moving the paretic limb in arc, while the affected arm remains attached to the body in semi-flexion [1]. The affected lower limb acts as if it were longer, being extended. Moreover, the ankle is internally rotated in clubfoot attitude (equinovarus), so all propulsion movement focuses on the hip. Thus, the hip and knee will be stiff and slightly flexed. As a result, the foot describes an arc foot internal concavity to go forward, touching the ground with the tip and the anterolateral face, tending to drag along the floor. The strong gluteal and quadriceps muscle groups are generally spastic and the most affected muscles are the iliopsoas (the strongest of the hip flexors), the hamstrings and the dorsiflexors of the foot. Moreover, almost always the posterior ones are better preserved than the anterior ones.

3.2.2. Gait analysis in hemiparesis

Hemiparesis is a common pathology in children. Therefore, there are several studies on the subject. For the professionals involved with patients with hemiparesis, observational gait analysis, linked with a thorough clinical evaluation, provides a tool for documenting gait deviations and can be used to determine causes of walking challenges. Physiotherapists use it on their daily basis. There are several predefined and implemented forms and codes for defining the degree of the pathology.

In the case of hemiparesis, there is a wide range of studies. Both observational and instrumented three-dimensional motion analyses are used to identify abnormal movement patterns during gait and underlying causes of deviations. Gait analysis provides an objective record of a child's gait before and after therapeutic intervention and should be considered a vital

part of the clinician's decision making [8].

During walking, the centre of body mass must pass from behind the weight bearing foot to in front of it. For this to take place, the foot must function as a sagittal plane pivot. Because the range required for this motion is approximately five times as great as both frontal and transverse plane motion, its evaluation becomes an essential part of a biomechanical assessment [12].

Gait abnormalities in children with cerebral palsy can affect movement at the hip, knee, and ankle. Therefore, a wide range of studies on hemiparesis focus on the parameters hip flexion, knee flexion, pelvic tilt, and ankle dorsiflexion of the sagittal plane. By looking at the patterns of this parameters, specific gait problems can be distinguished, due to deviations. For example, stiff-knee gait can be distinguished with the knee angle pattern. It is seen as an excessive knee flexion in stance continued into swing and a delayed and decreased knee flexion during swing.

For the purpose of this study and given the available facilities, the technique used for this study will be motion capture. Even though it gives three-dimensional information, only motion in sagittal plane will be studied.

3.3. The patients

Three patients with hemiparesis from Hospital Sant Joan de Déu were analysed at the Biomechanics Laboratory. The physiotherapist made a quick exploration on the patients, before proceeding with the movement captures. The relevant patient information for this project is listed in Table 1 (their real names have been changed):

Number	Patient	Age	Weight (kg)	Pathology	Notes
1	Maria	14	50,2	Right spastic hemiparesis	-
2	Jordi	15	55,0	Right spastic hemiparesis	He wears a Foot-Up
3	Isabel	11	61,8	Right spastic hemiparesis	She wears a DAFO

Table 1 Patient information

A Foot-Up (Fig. 6) is a lightweight ankle-foot orthosis by Össur that offers dynamic support for drop foot or similar complaints for which support of dorsiflexion is desirable. Note that Jordi did not wear the Foot-Up during the motion captures. Therefore, it did not affect the study.

A Dynamic Ankle-Foot Orthosis (DAFO) (Fig.7), by Cascade Dafo, Inc, is a brand name for a thin, flexible contoured brace, wrapping around the foot, which improves stability. Note that

Isabel wore a DAFO on her affected leg (right) during all the captures, since she could not walk without it.



Fig. 6 Foot-Up, by Össur



Fig. 7 DAFO, by Cascade Dafo, Inc.

4. Walking o'Clock device

4.1. Description

The walking o'Clock device is an electronic device with an inertial measurement unit (IMU) of nine axes (triple-axis accelerometers, gyroscopes and magnetometers). It works using quaternions to define the orientation of rigid bodies in three-dimensional space. It can record up to 200 measurements per second.

4.2. Functioning

Only the functioning of what was used in the project will be explained. Following the physiotherapist instructions, the device was placed on both thighs of the patients with a Velcro strap. Therefore, it was measuring the angle of the thigh with respect to the vertical axis in the sagittal plane. Since trunk orientation is vertical and almost constant during gait, it was considered that the device gave a relatively accurate estimation of the patients' hip flexion. Fig. 8 shows the device on the leg of patient Maria.



Fig. 8 On the left, the device on the leg of patient Maria. On the right, the device

First, with the patient standing straight, the device was set to angle 0° with the vertical axis to the ground. During the capture, via Wi-Fi, it would analyse the angles from the vertical axis and show their plot on screen. Then, as the capture stopped, it calculated the maximum and minimum angle. With that information, the physiotherapist decided what the best feedback could be.

The feedback can be sent through three ways: vibration, led or sound. Since some of the patients presented disturbances of sensitivity, the vibration was found as unreliable. Then, using earphones, the patient could hear the feedback.

The way the feedback works is the following. A threshold angle is chosen, which can be either of flexion or extension. Then, it can give feedback when the angle is under or when it is above the chosen one. To explain it, a reference Fig.9 has been made. If the chosen angle is θ_0 , there are two possible intervals: from 0° to θ_0 (called interval 1) and above θ_0 (interval 2). Thus, depending on the interval chosen, the patient will hear a sound either when they are under or above the angle θ . Taking the case of flexion, if the intention is for the patient to increase the angle, the feedback will be set on interval 1 and they will be told to keep flexing until the sound stops. On the other hand, if they need to decrease it, the feedback will be set on interval 2 and they will be told to stop flexing when they hear the sound. In the extension case, the angles will be negative, but the methodology would be the same one.

The feedback chosen for each patient during the capture motion is detailed in section 7.

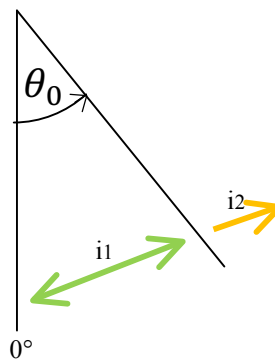


Fig. 9 Reference angle θ_0 to explain the functioning of the feedback in the case of flexion

5. Musculoskeletal model

5.1. Parameters of the musculoskeletal model

The study of human motion is based on the rigid body dynamics. The human body is defined as a multibody system, where each body segment, which is made of bone and soft tissues, is then assumed to be rigid. Thus, the inertial properties of each body can be defined through its mass, position of centre of mass and tensor of inertia. Depending on the intended application of the model, the complexity of it will vary. For example, simple non-muscle based models possess few variables, but they do not allow the study of muscle coordination during motion.

The system includes joint models that define kinematical constraints among the bodies and actuators that control them. These internal joints are modelled as rotations of one or more degrees of freedom. Moreover, the external joints must be defined, such as the contact of the feet with the ground. If there is not a fixed body, it is usual to consider that one of the system bodies is linked to the reference frame with a joint that allows the six degrees of freedom (three for rotation and three for translation). In this project, the pelvis will be linked to the ground (which defines the reference frame).

The number of bodies does not usually match the number of bones in a human body. In the case of a group of bodies that have a little relative movement among them, they may as well be considered as one body. In this project, since gait will be analysed, the head arms and trunk (HAT) are not considered in order to simplify the model. This practise is common when analysing the lower limbs during walking.

There are different options to model the joints. The first one is to try to get as close to reality as possible, by defining joints that only allow the rotations of human joints. This gives information about joint forces between bodies (the movements not allowed by the articulations are prevented by the called joint forces in classical mechanics). However, in biomechanics and specifically in motion capture, which are not exact and which throw unavoidable errors, this kind of forces could give resulting values that are far from reality.

There is the opposite option, which is to give more freedom to joints, by considering them as spherical joints and allowing small capture errors that will cause movements not allowed in reality. However, they would not be far from reality, since they would be the result of a real motion capture. This option also simplifies the difficulty when it comes to calculations.

The description of the configuration of the multibody system is done through the definition of a group of generalized coordinates associated to angles, and relative positions among the bodies. In this case, the generalized coordinates are the allowed joint rotations. Even though they will not be used in this project, the generalized velocities would be defined as time derivatives of the

generalized coordinates.

Regarding forces and moments, the actuators can be defined by directly linking them to all the generalized coordinates, so that each of the degrees of freedom is controlled by an actuator. In the case of human body modelling, as many angular actuators as relative rotations between bodies can be defined. For the ground joint, six actuators are defined (three linear and three angular), which are important for the analysis.

Being the system holonomic (the number of independent coordinates is the same as the number of degrees of freedom) and given the fact that its state is defined by a minimum number of generalized coordinates, the motion equations can be defined with Lagrange ordinary equations, as does the software OpenSim.

5.2. Model used in the project

The model used in the project is the Gait2392, provided by OpenSim. The model was created by Darryl Thelen (University of Wisconsin-Madison); and Ajay Seth, Frank C. Anderson and Scott L. Delp (Stanford University). Since the interest of the study was the lower body, the torso was removed from the model, along with its degrees of freedom. Moreover, the metatarsal joints of the model have been locked. This is due to the fact that they are not relevant to this study, and the patients may wear shoes, thus the information of the captures with the markers will not be enough to analyse the metatarsal joint.

The model that has been used for all the gait captures, even though it has been slightly modified when needed is the one explained below.

5.2.1. Bodies

The model is formed by 11 bodies and ground. To define each of the bodies, their mass, the position of its centre of inertia and the values of the elements of its central inertia tensor must be specified. The different bodies of the model are listed in Table 2 and shown in Fig. 10:

Name of the body	Name in OpenSim
Ground	<i>ground</i>
Pelvis	<i>pelvis</i>
Left femur	<i>femur_l</i>
Right femur	<i>femur_r</i>
Left tibia	<i>tibia_l</i>
Right tibia	<i>tibia_r</i>
Left talus	<i>talus_l</i>
Right talus	<i>talus_r</i>
Left calcaneus	<i>calcn_l</i>
Right calcaneus	<i>calcn_r</i>
Left phalanx bone	<i>toes_l</i>
Right phalanx bone	<i>toes_r</i>

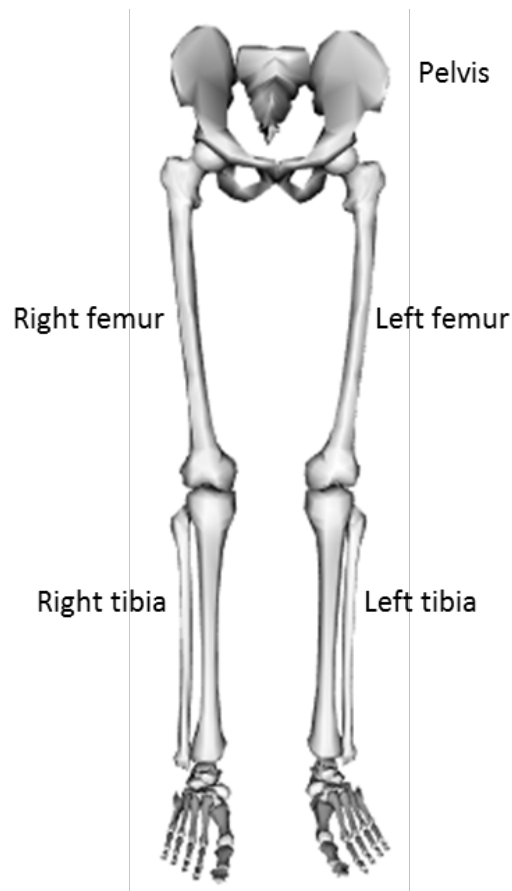


Table 2 Bodies of the multibody model

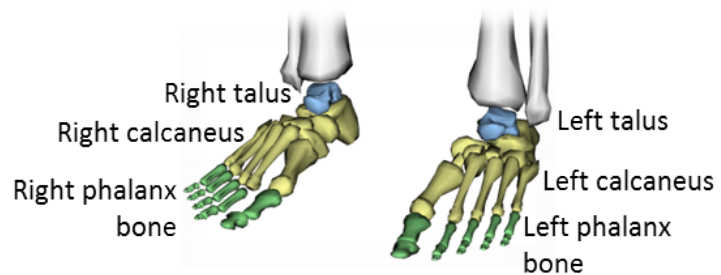


Fig. 10 Bodies of the model

5.2.2. Joints

The system has a total of 18 degrees of freedom (DOF), 6 for the pelvis motion with respect to the ground and the other 14 correspond to relative movements between the various bodies that form the model. However, as mentioned, the 2 degrees of freedom of the metatarsal joints are locked, thus the degrees of freedom disappear. Table 3 shows the existing joints, as well as the number of degrees of freedom allowed for each joint.

Name of the joint	DOF	Parent body	Son body	Name in OpenSim
Ground-Pelvis	6	Ground	Pelvis	<i>ground_pelvis</i>
Right hip	3	Pelvis	Right femur	<i>hip_r</i>
Right knee	1	Right femur	Right tibia	<i>knee_r</i>
Right ankle	1	Right tibia	Right talus	<i>ankle_r</i>
Right subtalar	1	Right talus	Right calcaneus	<i>subtalar_r</i>
Right metatarsal	0	Right calcaneus	Right phalanx bone	<i>mtp_r</i>
Left hip	3	Pelvis	Left femur	<i>hip_l</i>
Left knee	1	Left femur	Left tibia	<i>knee_l</i>
Left ankle	1	Left tibia	Left talus	<i>ankle_l</i>
Left subtalar	1	Left talus	Left calcaneus	<i>subtalar_l</i>
Left metatarsal	0	Left calcaneus	Left phalanx bone	<i>mtp_l</i>

Table 3 Model joints

5.2.3. Generalized coordinates

There are 20 generalized coordinates in the model, but those related to the metatarsal joint (*mtp_angle_l*, *mtp_angle_r*) are locked. Therefore, the system has 18 generalized coordinates, as expected, because the system is holonomic. These can be divided into two groups: the absolute coordinates (expressing configuration of the pelvis segment with respect to the ground) and the relative ones (expressing orientation between consecutive –parent and son– body segments). The coordinates classification appears in Table 4 Generalized coordinates, in which the first row indicates the son body and the parent one between brackets; and the rest of the rows show the coordinates expressing position and/or orientation of the son body with respect to the parent one.

Pelvis (Ground)	Femur (Pelvis)	Tibia (Femur)	Talus (Tibia)	Calcaneus (Talus)	Phalanx bone (Calcaneus)
<i>pelvis_tx</i>	<i>hip_adduction_l</i>	<i>knee_angle_l</i>	<i>ankle_angle_l</i>	<i>subtalar_angle_l</i>	<i>mtp_angle_l</i>
<i>pelvis_ty</i>	<i>hip_flexion_l</i>				
<i>pelvis_tz</i>	<i>hip_rotation_l</i>				
<i>pelvis_tilt</i>	<i>hip_adduction_r</i>	<i>knee_angle_r</i>	<i>ankle_angle_r</i>	<i>subtalar_angle_r</i>	<i>mtp_angle_r</i>
<i>pelvis_list</i>	<i>hip_flexion_r</i>				
<i>pelvis_rotation</i>	<i>hip_rotation_r</i>				

Table 4 Generalized coordinates

5.3. Marker protocol

5.3.1. The design of the protocol

After defining the model that will be used in the study, which has been taken from models made by experts on the matter, the marker protocol follows. Whereas the design of the model requires advanced knowledge in human physiology, the marker protocol design depends more on the motion capture.

In order to proceed with the design of the marker placement protocol, it is important to understand the aim of the motion capture taken, as well as the calculations involved.

The main analysis of this project will be inverse kinematics. This process, which will be explained in detail in section 8, obtains the time evolution of the generalized coordinates of the position of the markers over time, by solving an optimization problem that minimizes the error between the position of the experimental markers and the ones in the model. Therefore, it must be mathematically possible to perform inverse kinematics for the protocol to be valid. This means that it must be possible to determine the position and orientation of all the bodies of the biomechanical model at any instant, which depends on the mobility of joints. In general, to position body in the space, three markers are needed. For example, this is the case of the pelvis, which is a body linked to the ground with six degrees of freedom.

After knowing the position of the body with general movement, the position of its parent bodies must be determined. If the link with the parent is of three rotational degrees of freedom, two markers are needed to position the second body respect to the first, as long as none of them is in the common point between them. In case of having a joint with only one degree of freedom, a single marker will be enough, but always ensuring that this point does not belong to the reference of the parent body. In case of a relative movement of rotation with two degrees of freedom, one marker would be enough, as long as it is not in the axis of one of the rotations.

The protocol design must follow two criteria. The first one is to be as simple as possible, because having a greater amount of markers complicates the motion capture and the later data treatment. Secondly, more markers imply more information security, because, in case of losing a position of one of the markers at a time frame, the rest should be able to solve the inverse kinematics problem. Thus, it is important to keep both in mind when designing the protocol.

Another important point is the position of the marker in the body, since they must be easy to identify. Thus, it must be known anatomically speaking or by easily reaching it from other points. Moreover, they should be placed as close to the bone as possible, where there is least amount of tissue between the bone and the skin, in order to avoid soft tissue artefacts.

5.3.2. Protocol used in the project

The protocol used for all the captures is based on the Plug-in Gait marker placement, by ©Vicon Motion Systems, for the lower body. However, the thigh markers were deleted, due to the anatomical inaccuracy of placement. Moreover, the shank bone (tibia) marker was relocated at the tibial tuberosity below the knee. Finally, one marker was added on the toes. The protocol then includes 18 markers, which are distributed and grouped as follows in Table 5 and distributed as in Fig. 11, using the same names used in the OpenSim software:

Body	Markers in the body
Pelvis	R.ASIS L.ASIS R.PSI L.PSI
Right femur	R.GreatTroch R.Knee.Lat
Left femur	L.GreatTroch L.Knee.Lat
Right tibia (shank bone)	R. Shank.Front R.Ankle.Lat
Left tibia (shank bone)	L.Shank.Front L.Ankle.Lat
Right calcaneus	R.Heel
Right metacarpal	R.Toe.Lat
Left metacarpal	L.Toe.Lat
Right phalanx	R.Toe.Med
Left phalanx	L.Toe.Med

Table 5. Names of the markers used in the protocol

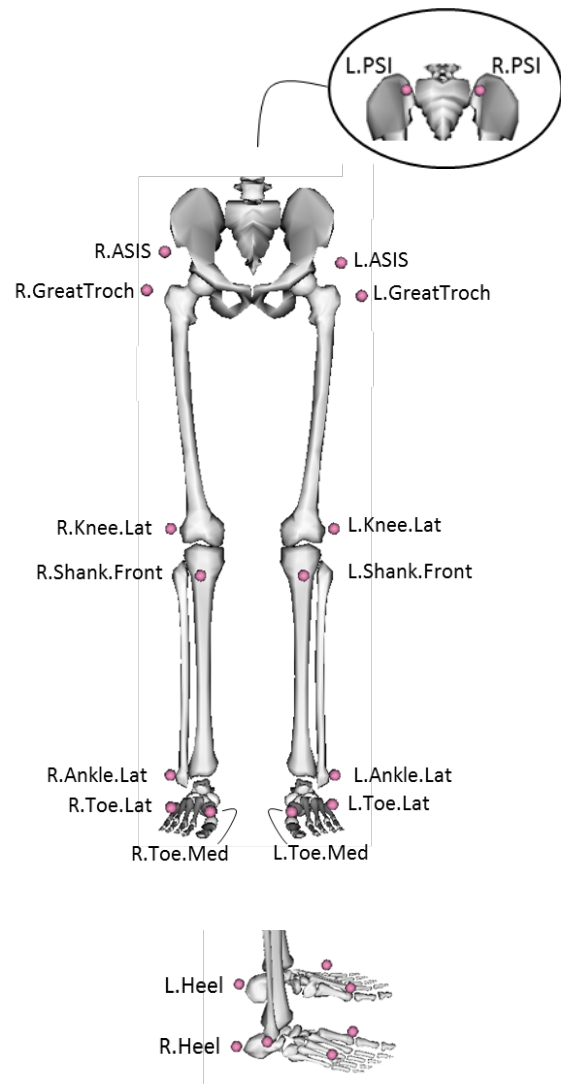


Fig. 11. Distribution of the markers in the body

6. Motion capture

6.1. The Biomechanics Laboratory

This sections presents the Biomechanics Laboratory of the UPC. It includes the description of the equipment, along with its characteristics. It also includes the calibration methodology, which is essential to ensure a precise motion capture.

6.1.1. Laboratory equipment

The UPC Biomechanics Laboratory, located in ETSEIB, involves a marker-based motion capture system OptiTrack™ from Natural Point, Inc.

It consists of 16 cameras, model V100:R2, equipped with IR (infrared light) LEDs. It is designed to capture the position of points within the capture space from the emission of infrared light, which is reflected in the markers (small spheres coated with a highly reflective fabric) fixed in the body being analysed. Therefore, the reflected light is discreetly captured by an optical system of cameras. Each of the cameras positions each marker in the perpendicular plane to its optical axis. The capture frequency is 100 Hz (100 images/second).

Using all the information captured by several cameras, the system is capable of computing the position of all the markers in the three-dimensional space. This information is processed by obtaining the position in space of all the markers present in the capture space. A marker and a camera can be seen in Fig. 12 and Fig. 13:



Fig. 12 OptiTrack Camera



Fig. 13 Marker

The cameras are arranged as shown in Fig. 14 A view of the Biomechanics Laboratory around the working space, eight located at three meters from the ground and six at a meter and a half.



Fig. 14 A view of the Biomechanics Laboratory

The signals from the cameras are transferred to a computer through the hubs, which are USB connection boxes where up to six cameras can be connected with USB 2.0 cables. These signals are processed with the software Motive, which allows not only the capture of the movement, but also the treatment of the captured data, as well as its export.

6.1.2. Calibration

In order to position the markers in the space, the camera system needs to be calibrated, which involves both dynamic and static calibration.

The dynamic calibration is achieved by waving a calibration wand (Fig. 15) with three markers attached to a pre-fabricated fixture in the work volume. Using the coincidental points and relative distances of the three markers and by capturing them, the intrinsic and extrinsic parameters of the camera system are calculated. The intrinsic parameters describe the variables that depend on the camera optics, whereas the extrinsic ones describe the spatial pose of the camera. While the wand is being moved, it is important to make sure that all the cameras are able to draw the path made with the wand, which can be seen on the computer screen. Each camera records in the plane perpendicular to its optical axis, thus in two dimensions. With the 2D images of all the cameras the three-dimensional image is created.

With the static calibration, the position and orientation of the camera with respect to the inertial coordinate system is determined. An L-shaped plate equipped with three markers set at a pre-defined distance is placed on the floor at the centre of the volume that the motion captures takes place, which is seen in Fig. 16.



Fig. 15 Calibration wand



Fig. 16 L-shaped plate with three markers

6.2. Procedure for motion capture

Given the fact that how cerebral palsy affects varies from one patient to another, even when having the same pathology (in this case, hemiparesis), this project is aimed to study each patient independently. However, a similar procedure has been followed to capture the movement of all subjects.

The capture of movement can be divided into four parts:

1. Static capture: it is used to scale the model when using the OpenSim software.
2. Natural gait (N): to show how the patient walks in their everyday life. It also allows the physiotherapist to evaluate how he should set the parameters of the device for the feedback.
3. Gait with feedback (F): after having the patient using the device for a while, it shows how it modifies their gait pattern. For patient Isabel, there will be two feedbacks (referred to as F1 and F2).
4. Gait right after the feedback (A): by stopping the feedback of the device, it shows if the gait pattern has changed compared to the natural one after using it for a while.

Since every patient is different, the feedback chosen for each case will vary. Moreover, in the case of one of the patients, Isabel, who has stability problems and more difficulty to walk, the gait after the feedback could not be analysed, due to her exhaustion. However, two different feedback cases for this case were taken, which are noted on.

As explained earlier when talking about the Walking O'Clock device, the physiotherapist chose the feedback that had to be set. In the table below, the feedback used in each patient is shown in Table 6:

Patient	Feedback
Maria	Flexion of 20° on right leg (F)
Jordi	Flexion of 25° on left leg (F)
Isabel	1. Extension of 0° on right leg (F1) 2. Flexion of 20° on right leg (F2)

Table 6. Feedback used in each patient

6.3. Motion capture and data export

6.3.1. Placement of the markers

In order to work with the data afterwards, the markers must be placed in the same anatomical points of the OpenSim model, following the protocol that has been explained in the previous section 5.3. The patients were asked to wear tight shorts, to prevent the markers from moving during the captures. They were placed on the shorts and skin of the patients with tape. Since the physiotherapist was there during the placement of the markers and he knew exactly how to find the anatomical points needed, he decided where to place them himself, which ensured more accuracy. Fig. 17 and Fig. 18 show the patients with the markers, as well as the device on patient Jordi.

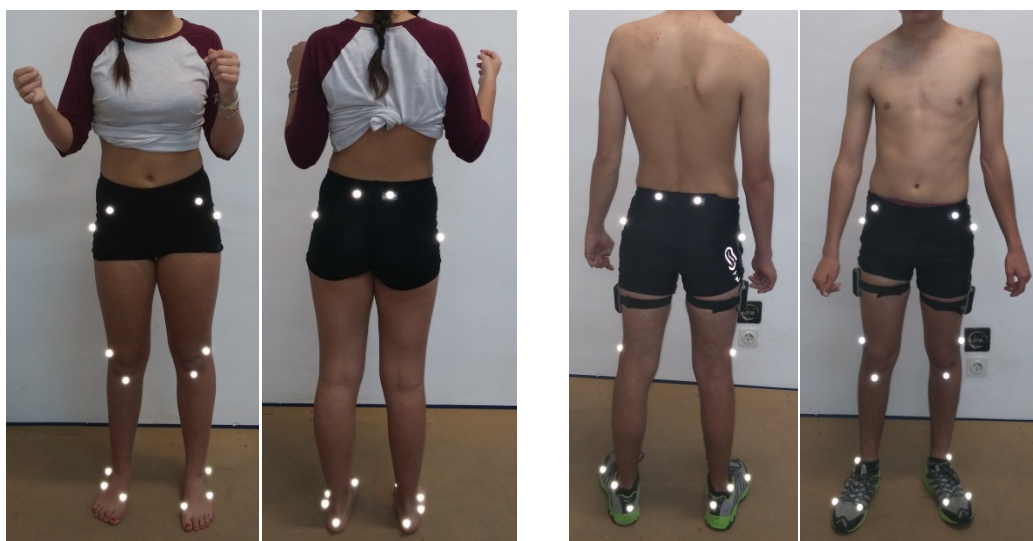


Fig. 17 On the left, patient Maria. On the right, patient Jordi, who can be seen wearing one device on each leg



Fig. 18 Patient Isabel

6.3.2. Optical tracking

Using the software OptiTrack Motive the position of the markers during the capture is tracked. Before starting the motion capture, the calibration file, which has been explained in section 3.2, must be loaded. Once the patient is standing in the working area, all the markers must be seen by the cameras. This is easy to check, because the software shows the number of markers that it is tracking and where they are, which is useful in case one of them falls. Also, it is important to remove any objects that may interfere with the cameras, such as jewellery or shiny materials. Once checked, the motion capture can start.

6.3.3. The walking path

Before starting the motion capture, a static pose capture was taken, which would be used for the data processing afterwards (in order to use it to scale the model in the software OpenSim, which is explained in 8.2). Then, each patient was asked to walk from one side of the Laboratory to the other side straight ahead, trying to do it in their regular speed. They could use the pressure plate as a visual reference, which were not used in this project, in order to keep a straight line. As seen in the left picture, there were two marks on the ground, which the patients could use to know where to start and turn to walk. In each walk from side to side, one capture was recorded, which it included, at least, three gait cycles. The same procedure was followed for every part for every patient. Fig. 19 shows patient Jordi going back and forth through the chosen path.

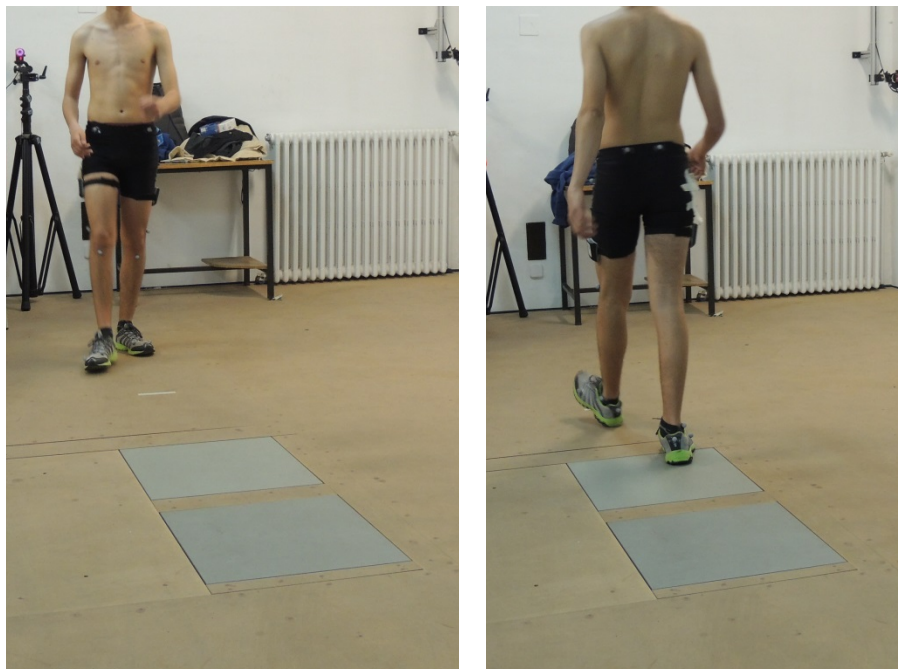


Fig. 19 Jordi path

6.3.4. Write the TRC file

In order to ensure a precise analysis of the gait pattern of every part, several captures were taken for each patient. For each of the parts (listed in section 6.2), three captures have been chosen, except from the static one, which was complete with only one capture. Therefore, ten captures per patient have been processed.

First of all, the captures had to be edited before being exported from the Laboratory Software, Motive. All the markers had to be named and checked. In case a marker was missing or had been lost by the program during a period of frames in the capture, the gap was filled with the editing tool fill gaps. Also, if the cameras lost a marker during the capture and it appeared as a new one, it was merged with the corresponding one with the editing tool merge markers. All of this had to be done manually and making sure not to miss any marker, or else the file would not be worked on with the software.

The files then were exported as a .csv file extension. That file would be then transformed into a .trc file, which is the one that the software OpenSim reads. To do so, a Matlab program was created and used, which directly converted the files into .trc, just by executing them. Basically, what it did was:

1. Read the .csv file and create one with Matlab extension
2. Read all number of the file, along with the information about rows and columns
3. Read how many markers would be in the capture and the corresponding names
4. Adjust the axes:

`x_OpenSim=z_cameras`

`y_OpenSim=y_cameras`

`z_OpenSim=-x_cameras`

The files are then ready to be opened with the OpenSim software.

7. Kinematics analysis

7.1. Introduction to OpenSim

This section explains the main options of OpenSim 3.1, which is the program that has been used to perform kinematic calculations. Once the program has been opened and the model loaded, the interface looks as seen in Fig. 20:

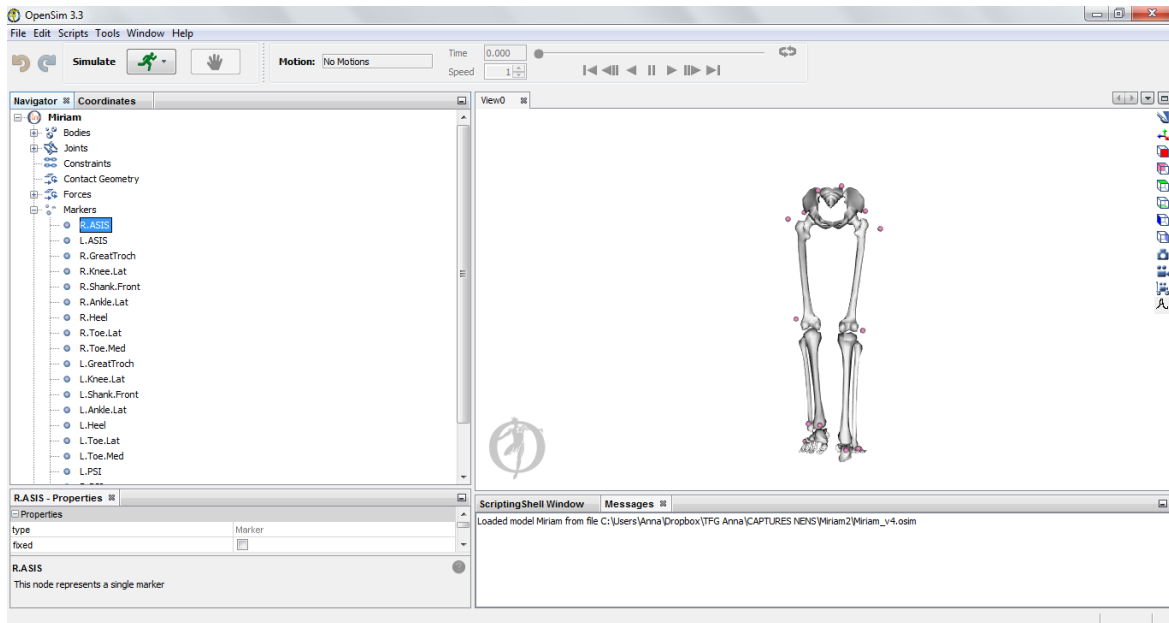


Fig. 20 OpenSim user interface

The browser (Navigator) allows choosing which elements are displayed, having the options of Bodies, Joints or Markers. Moreover, by clicking the right mouse button, several display actions are unfolded. More than one model can be added in the browser, but only one can be active, which is achieved by clicking on the Make Current option.

All the tools for the analysis of motion are included on the Tools menu. The ones used in this project are the following:

- Scale model: based on a generic model skeleton, it transforms its dimensions to those of the subject of the capture.
- Inverse Kinematics: it calculated the value of the generalized coordinates over time from the marker position measurements.

7.2. Scaling the model

The first step for the motion analysis is to scale the model, in order to create a skeleton model as close as possible to the real subject. In addition, it allows relocating the markers in the model in most similar way to how they were placed in reality. It is advisable to relocate them (Adjust Markers Model) once the model has already been adjusted and, therefore, the size is the appropriate one.

For this process, two files are necessary: the file with the generic model that will be modified (.osim) and the file motion capture that will enable the changes (.trc). The result of the scaling is a scaled new model, with the extension .osim, that can be saved for later use.

The process starts by going on the Tools menu > Scale Model. The first step is to specify the mass of the model, which is adjusted for each subject according to their total mass. The mass is needed for the analysis of dynamics. Thus, it is not essential in this project, since it will only take into account the analysis of kinematics, but, this way, the model is adjusted and scaled properly.

This mass will be distributed segment by segment in proportion to the original model. Therefore, in the Scale Model section, the option Preserve mass distribution during scale must be active and the direction of the .trc file of the static pose must be indicated. In the first scale, the Adjust Model Markers must be deactivated, to only change the dimensions of the skeleton and not the position of the sensors in the model.

The second tab of the tool has the option Edit Measurement Set, which is needed to define the scale factors, which are the relationship of the distance between two markers in reality and in the model. Once these factors are defined, they must be associated to the solids of the model. Fig. 21 shows how this factors are defined, which is by following (Eq 1:

$$scale_factor_1 = \frac{e_1}{m_1} \quad scale_factor_2 = \frac{e_2}{m_2} \quad (\text{Eq 1.})$$

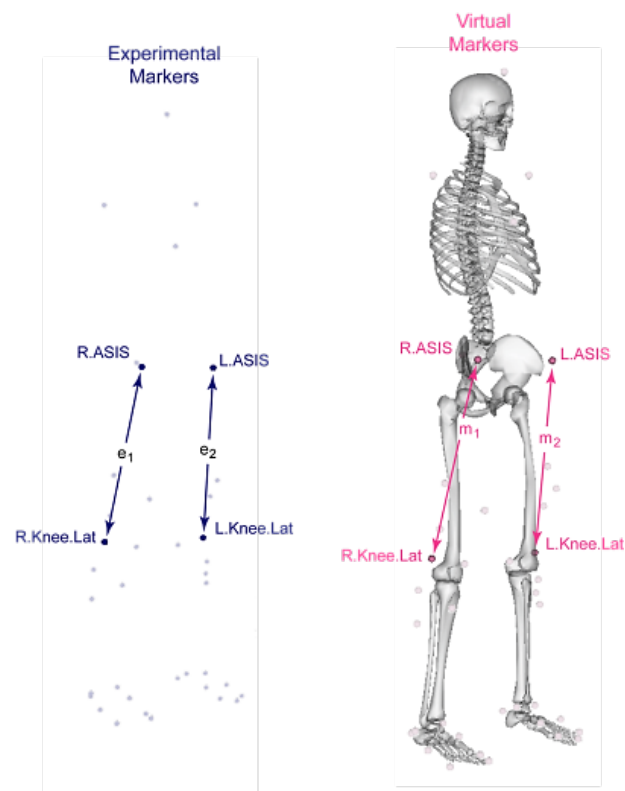


Fig. 21 Distance of two markers. On the left, experimental markers. On the right, markers from the model. Taken from the OpenSim guide [2].

There is the option to scale the segments with the same factor in all three directions or discriminate different factors in different directions. This depends on the markers used and how they can be related. Fig. 22 shows the markers pairs for the measurements used to apply the scale factors. Fig. 23 shows the applied scale factors that OpenSim has then calculated.

Measurements	Marker Pairs						
✖ pelvis_trans	+	L.PSI	R.PSI	✖	R.ASIS	L.ASIS	✖
✖ pelvis_long	+	R.PSI	R.ASIS	✖	L.PSI	L.ASIS	✖
✖ pelvis_vert	+	R.ASIS	R.GreatTroch	✖	L.ASIS	L.GreatTroch	✖
✖ R.Thigh	+	R.GreatTroch	R.Knee.Lat	✖			
✖ L.Thigh	+	L.GreatTroch	L.Knee.Lat	✖			
✖ R.Shank	+	R.Knee.Lat	R.Ankle.Lat	✖			
✖ L.Shank	+	L.Knee.Lat	L.Ankle.Lat	✖			
✖ R.Calc_long	+	R.Heel	R.Toe.Lat	✖			
✖ L.Calc_long	+	L.Heel	L.Toe.Lat	✖			
✖ R.Toes_trans	+	R.Toe.Lat	R.Toe.Med	✖			
✖ R.Toes_long	+	R.Toe.Lat	R.Heel	✖			
✖ L.Toes_trans	+	L.Toe.Lat	L.Toe.Med	✖			
✖ L.Toes_long	+	L.Toe.Lat	L.Heel	✖			

Fig. 22 Marker pairs for the measurements used to scale

Body Name	Measurement(s) Used			Applied Scale Factor(s)		
ground	Unassigned					1.0
pelvis	pelvis_long	pelvis_vert	pelvis_trans	1.068848	0.773113	1.051243
femur_r	R.Thigh					0.769209
tibia_r	R.Shank					0.926490
talus_r	R.Calc_long					1.147271
calc_n_r	R.Calc_long	R.Toes_trans	R.Toes_trans	1.147271	1.026652	1.026652
toes_r	R.Toes_long	R.Toes_trans	R.Toes_trans	1.147271	1.026652	1.026652
femur_l	L.Thigh					0.811810
tibia_l	L.Shank					0.899832
talus_l	L.Calc_long					1.144999
calc_n_l	L.Calc_long	L.Toes_trans	L.Toes_trans	1.144999	1.001410	1.001410
toes_l	L.Toes_long	L.Toes_trans	L.Toes_trans	1.144999	1.001410	1.001410

Fig. 23 The applied scale factors for each body and the measurements used

In Fig. 24, the changes in the model after the scale are noticed, since the patient is shorter, her pelvis is wider and, thus the legs are further apart.

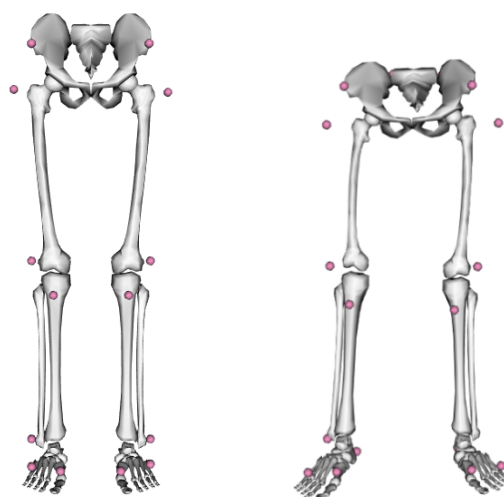


Fig. 24 On the left, the generic model. On the right, the scaled model.

After the skeletal model scale is done, the option *Adjust Model Markers* follows, to relocate the markers as similar to where they were in reality as possible from a .trc file. In the *Static Pose Weights*, it is desirable to set different weights on the markers, since this is the weight that each marker will have in the optimization carried to adjust its position. Low weight markers are those with less precise positions, due to being away from the bone and with the possibility of soft tissue movements. On the other hand, high weight markers are those with a precise anatomic location. The chosen weights can be seen in Fig. 25:

Settings Scale Factors Static Pose Weights

☐ Enable all selected Value ☐ From file Weight

☐ Disable all selected ☐ Default value

☐ Manual value

Enabled	Marker Name	Value	Weight
<input checked="" type="checkbox"/>	R.ASIS	From File	10.0
<input checked="" type="checkbox"/>	L.ASIS	From File	10.0
<input checked="" type="checkbox"/>	R.GreatTroch	From File	1.0
<input checked="" type="checkbox"/>	R.Knee.Lat	From File	10.0
<input checked="" type="checkbox"/>	R.Shank.Front	From File	0.0
<input checked="" type="checkbox"/>	R.Ankle.Lat	From File	1.0
<input checked="" type="checkbox"/>	R.Heel	From File	1.0
<input checked="" type="checkbox"/>	R.Toe.Lat	From File	1.0
<input checked="" type="checkbox"/>	R.Toe.Med	From File	1.0
<input checked="" type="checkbox"/>	L.GreatTroch	From File	1.0
<input checked="" type="checkbox"/>	L.Knee.Lat	From File	10.0
<input checked="" type="checkbox"/>	L.Shank.Front	From File	0.0
<input checked="" type="checkbox"/>	L.Ankle.Lat	From File	1.0
<input checked="" type="checkbox"/>	L.Heel	From File	1.0
<input checked="" type="checkbox"/>	L.Toe.Lat	From File	1.0
<input checked="" type="checkbox"/>	L.Toe.Med	From File	1.0
<input checked="" type="checkbox"/>	L.PSI	From File	10.0
<input checked="" type="checkbox"/>	R.PSI	From File	10.0

Enabled	Coordinate Name	Value	Weight
---------	-----------------	-------	--------

Fig. 25 Static Pose Weights for each marker

To check whether the scale process has been successful, the experimental markers of the *static pose* can superimpose the model. To do so, inverse kinematics of the *static pose* must be done first for the scaled model (this process will be in detail explained further). Once done, on the section Motions of the OpenSim browser a file named Results appears. This file must be linked to the experimental markers, by clicking on the Results file with the right mouse button and selecting Associate Motion Data, adding then the .trc file of the *static pose*. Fig. 26 shows this association, where the two markers can be seen and it is visible that the scale process was satisfactory:



Fig. 26 On the left, the scaled lower body model, with model (pink) and experimental (blue) markers. On the right, a closer view of the feet.

7.3. Inverse kinematics

Once the model is scaled and adjusted for the patient that is being analysed, the kinematics of the system bodies can be obtained through inverse kinematics. This tool, which is called *Inverse Kinematics* in OpenSim, uses the .trc file of the motion capture to obtain the evolution of the generalized coordinates over time. The resulting file is a *Motion* file (.mot extension), which contains the value of the generalized coordinates of the model in every instant of the capture.

7.3.1. Calculation of inverse kinematics

Inverse kinematics solve an optimization problem, based on minimizing the distance between the position of the markers of the OpenSim skeletal model and the experimental markers placed in the body of the patient for every instant of time. Thus, the objective function to be minimized is (Eq 2. :

$$\min_q \left[\sum_{i=1}^m w_i \|x_i^{exp} - x_i(q)\|^2 \right] \quad (\text{Eq 2.})$$

Being $x_i(q)$ the position vector of the i th marker of the OpenSim skeleton model, which depends on the system configuration described by the vector of the generalized coordinates q . Moreover, x_i^{exp} is the position vector of the i th experimental marker (placed in the body of the patient). Finally, w_i is the weight assigned to the i th marker. As mentioned, this weight is the degree of importance given to the markers, depending on how clear their position in the body is. Thus, the markers with more weight will be those whose location is more precisely defined, whereas the others will be the ones with less clear points anatomically speaking or that can have more soft tissue movement (i.e., those that are away from the bone).

7.3.2. Solution with OpenSim

As presented at the beginning of this section, all the tools for the analysis of motion are included on the *Tools* menu. Thus, it is found in *Tools> Inverse Kinematics*. In order to run it, the model must be active. In the first tab of the tool interface, the markers file direction must be indicated (Marker data for trial), as well as the time lapse in which you want to find a solution. In the second tab, the weight can be specified. This diagram in Fig. 27 shows all the files used in inverse kinematics:

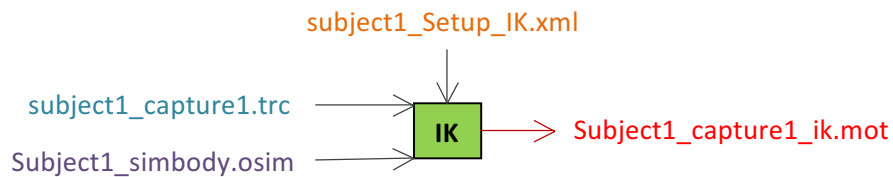


Fig. 27 Files involved in inverse kinematics (IK)

The result is a .mot file with as many rows as frames had the captures, with the defined coordinates of the model as columns. This result must be saved by clicking on the right button of the *Result* file that appears on the *Motions* section of the browser.

When the kinematics is done, the *Messages* window of OpenSim shows the report of the results, which evaluates the marker errors. It does it by finding the quadratic error between the position of each experimental marker and its position in the model configuration resulting from the inverse kinematics (Eq 3. :

$$e_{mark} = \|x_i^{exp} - x_i(q)\|^2 \quad (\text{Eq 3.})$$

The errors can also be evaluated visually, by associating the file to the motion capture file, so it shows the movement of the model and of the markers simultaneously. Moreover, OpenSim can show the error of every marker for every frame. The maximum error should not exceed 2-4 cm and the RMS value of the error cannot be greater than 2 cm, for all the markers.

8. Results and discussion

This chapter presents the results obtained from the motion capture, and an analysis of these data. It first starts with the methodology followed to treat all the data obtained from the software OpenSim. Afterwards, the four generalized coordinates of the sagittal plane will be analysed for each patient.

8.1. Methodology

After executing the inverse kinematics in the OpenSim software, the generalized coordinates of each capture were exported to a .mot file. In order to read this file, a Matlab program was created. Since the objective of this project is to analyse how the gait pattern changes when using the *Walking o'Clock* device, one gait cycle of each capture will be analysed. The methodology followed for the three patients will be explained in this section.

First of all, when any of the captures was exported from the Motive software, as explained in section 7, it had a number of frames, which is equivalent to the time that the capture had been recorded. Since every capture had at least three complete gait cycles, it was essential to know from which frame to which frame the wanted gait cycle goes, in order to identify it from the whole motion capture. As mentioned in 4.3.2 with more detail, the gait cycle starts when one foot touches the ground (IC), and the cycle finishes when the same foot touches de ground again. In the study, the right foot was taken as the reference. Therefore, when the subject stepped on his/her right foot, the gait cycle started, ending with the same right foot stepping back on the ground. To decide which frames are included in the chosen gait cycle, the y axis of the right heel was plotted on Matlab. This way, the frames could be chosen by looking at when the y axis of the right heel marker reached the minimum value. The information of the marker was imported from the .csv file taken from the Motive software, which is read in Matlab.

Fig. 28 shows how the frames of the gait cycle for the first natural motion capture of patient Maria were defined. The cursors show that it starts at frame 63 and ends at frame 181. The same procedure is followed for all the other captures. These frames will be used in all the programs created to analyse every capture.

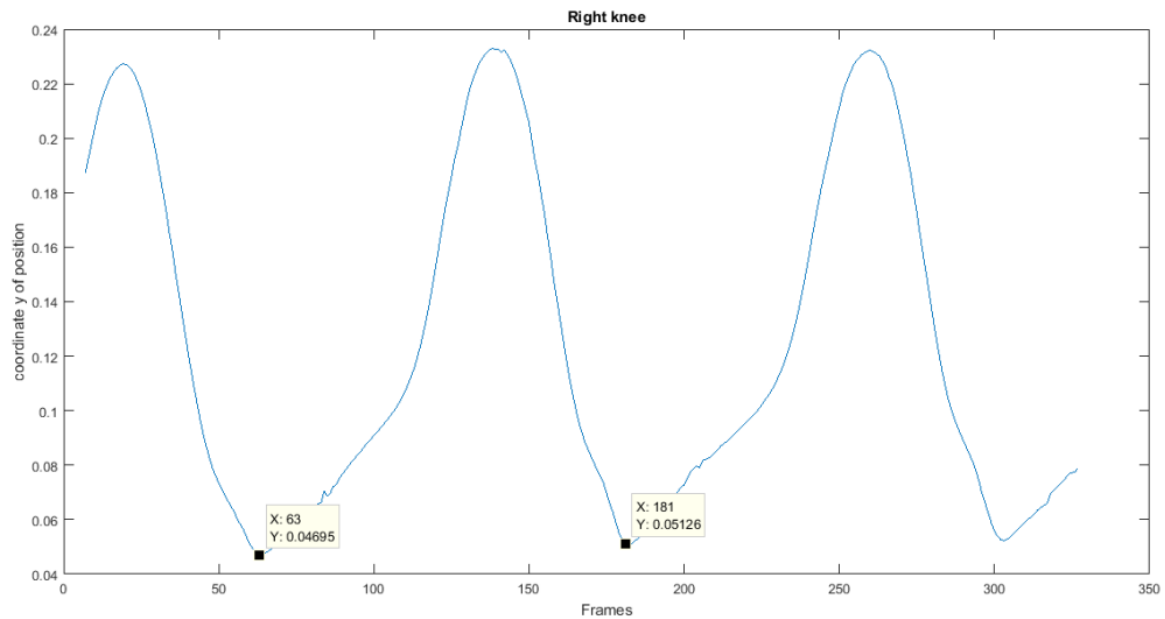


Fig. 28 The plot that shows the coordinate y of the position of the right foot with the frame. N1, Maria

The following Table 7 shows the chosen frames for all the captures. On the first column appears the name of the capture, on the second one the initial frame (IF) and on the third the final frame (FF). The names of the captures are the ones defined in 7.1

Patient								
Maria			Jordi			Isabel		
Capture	IF	FF	Capture	IF	FF	Capture	IF	FF
N1	61	177	N1	160	275	N1	401	588
N2	194	310	N2	116	228	N2	274	488
N3	173	290	N3	74	177	N3	378	558
F1	283	410	F1	158	269	F11	585	786
F2	109	214	F2	123	229	F12	153	327
F3	45	155	F3	176	285	F13	677	839
A1	43	155	A1	209	325	F21	425	580
A2	89	207	A2	170	283	F22	618	798
A3	208	327	A3	163	279	F23	410	596

Table 7. Initial and final frames of the analysed gait cycles for each motion capture.

Having the frames, the methodology followed with all the gait cycles was the same for the three patients, but always separately. As commented, this project studies the kinematics of gait and only the angular coordinates in the sagittal plane will be analysed. These are:

- Pelvic tilt angle
- Right and left knee angles
- Right and left hip angles
- Right and left ankle angles

The analysis steps followed are the following:

1. Plotting the coordinates of each gait cycle and checking that all the values are logical
2. Calculating the maximum and minimum angles of each gait cycle. Since the physiotherapist took the maximum and minimum angles of the thigh (considered as hip flexion angles as explained in section 4) to think of the best feedback for the patient, it is interesting to compare them.
 - a. For each phase, find the mean value for the maximum and minimum angle (3 per patient)
 - b. With the mean values, find the ROM of each phase and the percentage that has changed from the natural gait
3. Finding the equation that describes each coordinate for each gait cycle and generate the mean for each phase. Then, for each coordinate, three plots are expected (for example, in the case of patient Maria, one for the natural phase, one for the feedback phase and one for the after feedback phase)
4. To calculate the Root Mean Square Error (RMSE) comparing the three phases.
 - a. For patients Maria and Jordi, three RMSE will be calculated:
 - i. Natural pattern vs Feedback
 - ii. Natural pattern vs After Feedback
 - iii. Feedback vs After Feedback
 - b. For patient Isabel two RMSE will be calculated:
 - i. Natural pattern vs Feedback 1
 - ii. Natural pattern vs Feedback 2
5. To merge the plots of the three phases and compare them.

There are different facts that are important to mention. First of all, when proceeding with the inverse kinematics, some abnormalities were seen, since the pelvis took a position impossible

to reach in reality (completely twisted). After having investigated the captures that presented this problem, the cause was found as being a problem of the direction of the walk. As explained in section 7.2.3, the patients first walked to one side and then came back. Therefore, when they came back, since they rotated the pelvis completely, the model got confused. This was solved by directly changing the model and broadening the limit values of these coordinates.

After the inverse kinematics, when calculating the maximum and minimum angles, another abnormality appeared in some cases, such as angles that could not be possible in reality. Therefore, by looking back at the performance of the inverse kinematics tool in OpenSim, the problem was visible. By checking the captures that had problems with the angles, it could be seen that they reached limit values in the coordinates of OpenSim. This was also solved by changing the limit values for the coordinates that gave problems.

It is important to keep in mind that the constraints given to the generalized coordinates only affect the dynamics of the capture, not the inverse kinematics, so the values can be changed as wanted with no consequences in this case.

This problem may be due to that, when the capture was taken, the patient may have not walked perfectly straight. Also, given the fact that the patients may present deformities in their skeleton, OpenSim may get confused. In addition, there were a couple of captures that kept giving wrong angles, even having changed the model. In these cases, the solution was to go back to the Motive software and export new captures of the same phase in substitution of the abnormal one. For these new captures, the same methodology had to be followed.

Another problem that appeared in some captures was that, after analysing them, an odd behaviour was seen, compared to the other captures of the same gait, such as unexpected peaks. Therefore, after trying to understand the cause of such behaviour without success, the solution taken was to change the gait cycle chosen from that capture, by taking the one before or after (changing then the frames), so that the capture itself was not changed.

As mentioned before, the three patients will be analysed separately. The objective is to compare the three different phases for each of them. Moreover, the shape of the plots of the studies coordinates will be visually compared to the standard ones, presented in 4.3.2. However, since it is a fact that the patients have a pathology that modifies their gait pattern, it is expected that it will vary from the standard one. Then, it may allow seeing if the affected leg approaches the standard pattern with the use of the device. For each patient, the four above-mentioned coordinates will be analysed by analysing the maximum and minimum angles, the motion range and the graphs representing the three phases, along with the RMSE.

8.2. Maria

Patient Maria was put under a feedback of 20° angle of the right thigh (considered as hip flexion).

8.2.1. Hip flexion

Fig. 29 and Fig. 30 show the hip flexion angle pattern for both legs in a GC:

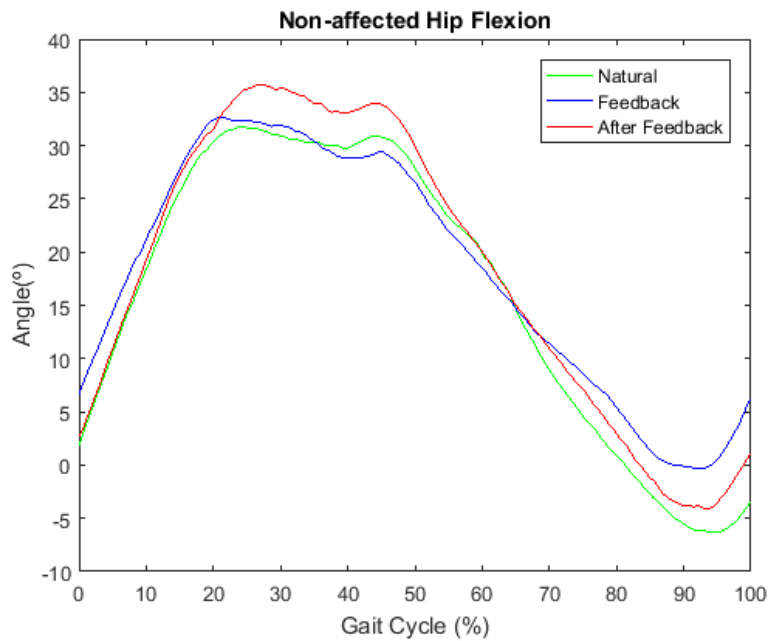


Fig. 29. Non-affected hip flexion angle in a GC

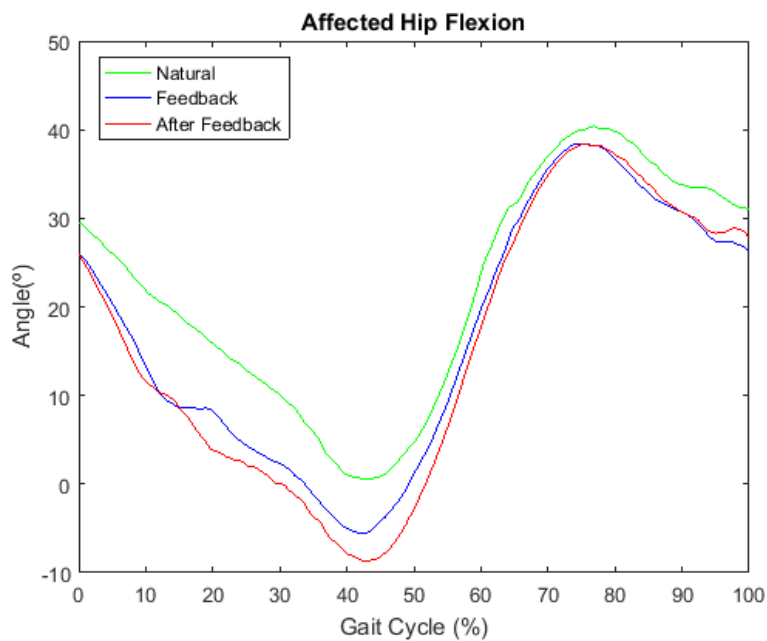


Fig. 30. Affected hip flexion angle in a GC

Coordinate	Angle type	Natural	Feedback	After feedback
Affected hip flexion	max	40,39	38,37	38,37
	min	0,64	-5,63	-8,64
Non-affected hip flexion	max	31,81	32,63	35,76
	min	-6,31	-0,32	-4,12

Table 8. Maximum and minimum angles of hip flexion in a GC

Coordinate	Natural	Feedback	After feedback	%F/N	%A/N
Affected hip flexion	39,75	44,00	47,01	10,69	18,27
Non-affected hip flexion	38,12	32,95	39,88	-13,57	4,62

Table 9. Hip ROM

Coordinate	Natural-Feedback	Natural-After feedback	Feedback-After feedback
Affected hip flexion	5,70	7,31	2,12
Non-affected hip flexion	3,43	2,39	2,97

Table 10. RMSE between phases

As seen in Table 8, the maximum angle for the affected and non-affected hip flexion has not had an important variation with the feedback. The difference appears when it comes to minimum angles for both legs. Whereas the minimum angle is greater (in absolute value) for the affected hip, it decreases for the non-affected one during the feedback.

The ROM for the hip of the affected limb has increased in both cases (F and A, see Table 9). It is interesting to notice that, during the feedback, while the affected hip increased its ROM, the non-affected one decreased it. When looking at the ROM for both hips during feedback, the difference is of more than 10°. In this case, the patient moved more the affected hip.

When looking at the pattern of all the captures, it is visible that the affected limb is the one that suffers most change with respect to the natural motion. This is also backed up with the RMSE

values (Table 10) when comparing the natural gait to the feedback and after feedback ones. What is unexpected is that the difference is greater for the after feedback gait, because it was thought that the patient would tend to go back to the natural pattern in after feedback. Moreover, notice the low values of the RMSE between feedback and after feedback, which show that the patient kept a similar gait pattern after stopping the device.

8.2.2. Pelvic tilt

The pattern of the pelvic tilt angle in a GC is seen in Fig. 31:

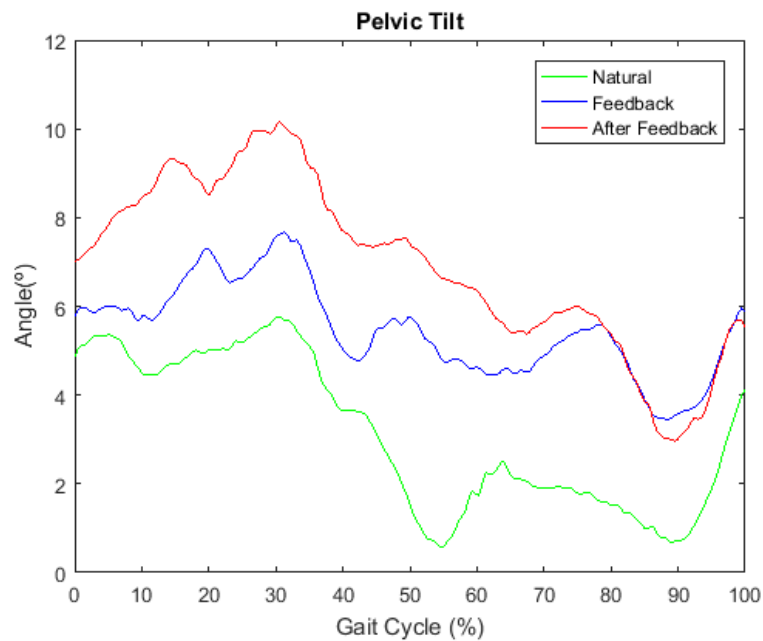


Fig. 31. Pelvic Tilt angle in a GC

Coordinate	Angle type	Natural	Feedback	After feedback
Pelvic tilt	max	5,75	7,68	10,16
	min	0,57	3,45	2,97

Table 11. Maximum and minimum angles of pelvic tilt in a GC

Coordinate	Natural	Feedback	After feedback	%F/N	%A/N
Pelvic tilt	5,18	4,23	7,19	-18,31	38,77

Table 12. Pelvic tilt ROM

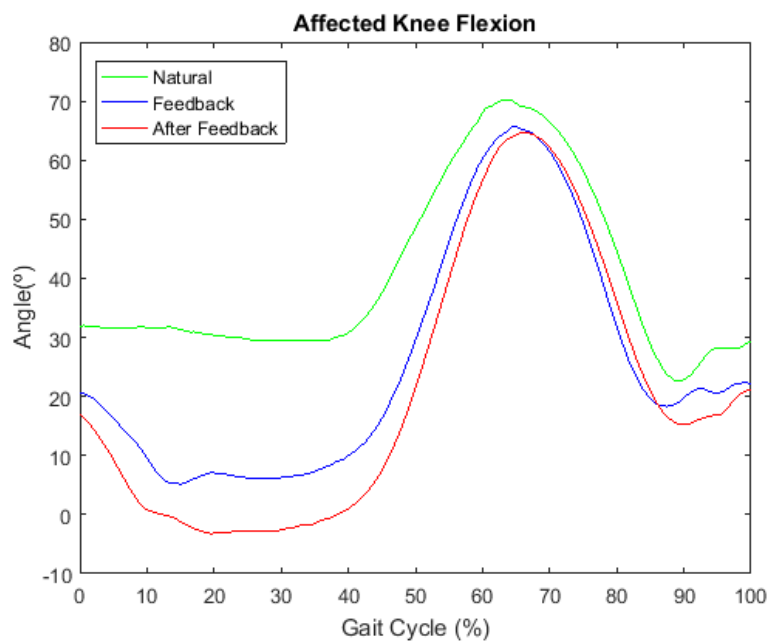
Coordinate	Natural-Feedback	Natural-After feedback	Feedback-After feedback
Pelvic tilt	2,56	3,95	1,81

Table 13. RMSE between phases

Notice that the maximum angles increase in both feedbacks (Table 11). However, the minimum angles also increase in value, shifting the pattern upwards, as seen in Fig. 31. During feedback, the pelvic tilt pattern is closer to the standard one, which is supported by the lower ROM (Table 12). This fact indicates that the patient gains stability when using the device. However, in after-feedback, ROM is higher than in natural gait, so the patient loses stability. This may be due to the transition between states and the confusion that may have caused her. As for hip flexion, the RMSE value between feedback and after feedback is low (Table 13).

8.2.3. Knee flexion

Fig. 32 and Fig. 33 show the knee flexion angle pattern in a GC for both legs:

**Fig. 32. Affected knee flexion angle in a GC**

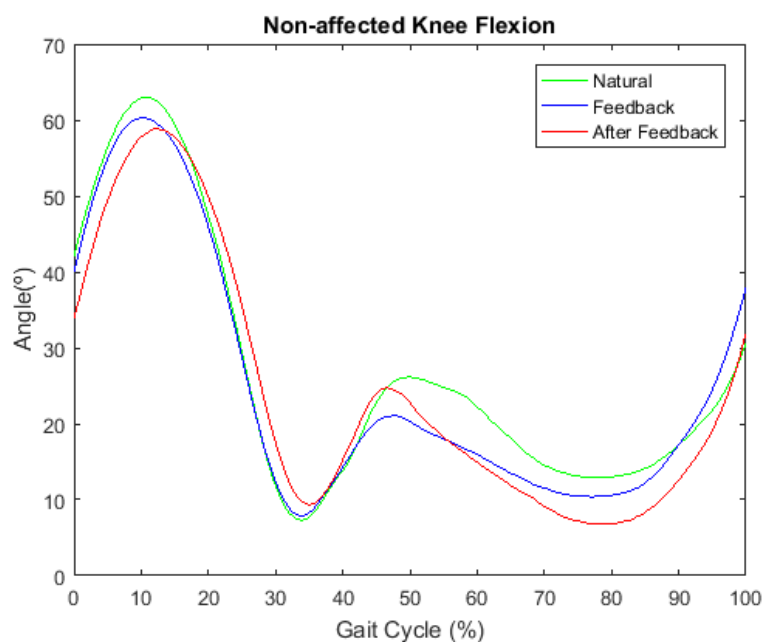


Fig. 33. Non-affected knee flexion angle in a GC

Coordinate	Angle type	Natural	Feedback	After feedback
Affected knee flexion	max	70,22	65,85	64,69
	min	22,62	5,20	-3,20
Non-affected knee flexion	max	62,97	60,26	58,84
	min	7,33	7,89	6,75

Table 14. Maximum and minimum angles of knee flexion in a GC

Coordinate	Natural (N)	Feedback (F)	After feedback (A)	%F/N	%A/N
Affected knee flexion	47,60	60,65	67,89	27,42	42,63
Non-affected knee flexion	55,64	52,37	52,10	-5,88	-6,37

Table 15. ROM of the knee flexion angle

Coordinate	Natural-Feedback	Natural-After feedback	Feedback-After feedback
Affected knee flexion	16,60	22,57	6,49
Non-affected knee flexion	3,36	5,02	3,80

Table 16. RMSE between phases

There is a negative angle in after-feedback for the affected knee during stance (Table 14), which means that there is genu *recurvatum*, also referred to as hyperextension [8].

While the knee ROM in the natural pattern is lower for the affected leg (Table 15), it becomes greater with the feedback. The difference in ROM varies dramatically when using the device. As seen in the graph above, it is obvious that this change occurs during the first phases of the gait cycle, when the angle highly decreases. The non-affected limb does not present any important changes when it comes to maximum and minimum angles, even though the ROM slightly decreases with the feedback.

The most important difference of the knee flexion is for the affected limb. As easily seen in the graphs, the affected knee angle presents a flat tendency during the first phases of gait, which is known as stiff knee. As seen, the stiffness does not disappear when using the device, but it is clearly modified and not as flat. Actually, the after feedback pattern resembles more the normal shape than the other two. The RMSE values show the great difference between patterns for the affected limb (Table 16).

8.2.4. Ankle dorsiflexion

In Fig. 34 and Fig. 35, the angle pattern for ankle dorsiflexion for both legs in a GC is seen:

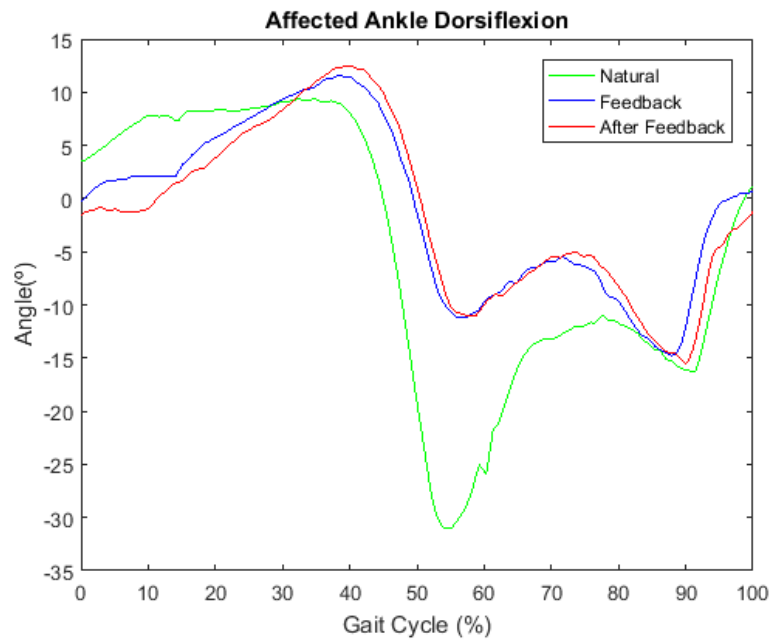


Fig. 34. Affected ankle dorsiflexion angle in a GC

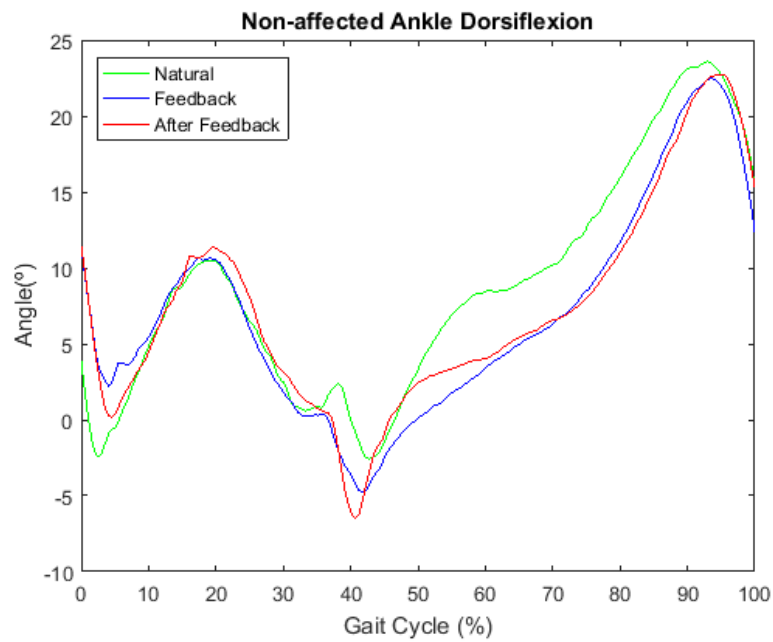


Fig. 35. Non-affected ankle dorsiflexion angle in a GC

Coordinate	Angle type	Natural	Feedback	After feedback
Affected ankle dorsiflexion	max	9,37	11,55	12,51
	min	-31,05	-14,76	-15,50
Non-affected ankle dorsiflexion	max	23,59	22,47	22,74
	min	-2,58	-4,77	-6,48

Table 17. Maximum and minimum angles of ankle dorsiflexion in a GC

Coordinate	Natural (N)	Feedback (F)	After feedback (A)	%F/N	%A/N
Affected ankle dorsiflexion	40,42	26,31	28,01	-34,91	-30,71
Non-affected ankle dorsiflexion	26,17	27,24	29,22	4,09	11,67

Table 18. ROM of the ankle dorsiflexion angle

Coordinate	Natural-Feedback	Natural-After feedback	Feedback-After feedback
Affected ankle dorsiflexion	8,21	8,89	1,94
Non-affected ankle dorsiflexion	3,25	3,13	1,26

Table 19. RMSE between phases

The affected ankle, which has a larger minimum angle (in absolute value) in the natural gait, is the most modified. The ROM greatly decreases and its value approaches the one of the non-affected ankle.

Satisfactorily, the pattern of feedback and after feedback for the affected side is more similar to the standard one. The abnormal peak that appeared in pre-swing of the GC disappears when using the device, which shows more control of the ankle.

Also, an abnormal peak appears in the natural gait of the non-affected ankle during the terminal swing of the GC, which softens with feedback. However, at the same point, a pronounced low peak appears in the after feedback.

8.3. Jordi

Patient Jordi was put under a feedback of 25° angle of the left thigh (considered as hip flexion).

8.3.1. Hip flexion

Fig. 36 and Fig. 37 show the hip flexion angle pattern for both legs in a GC:

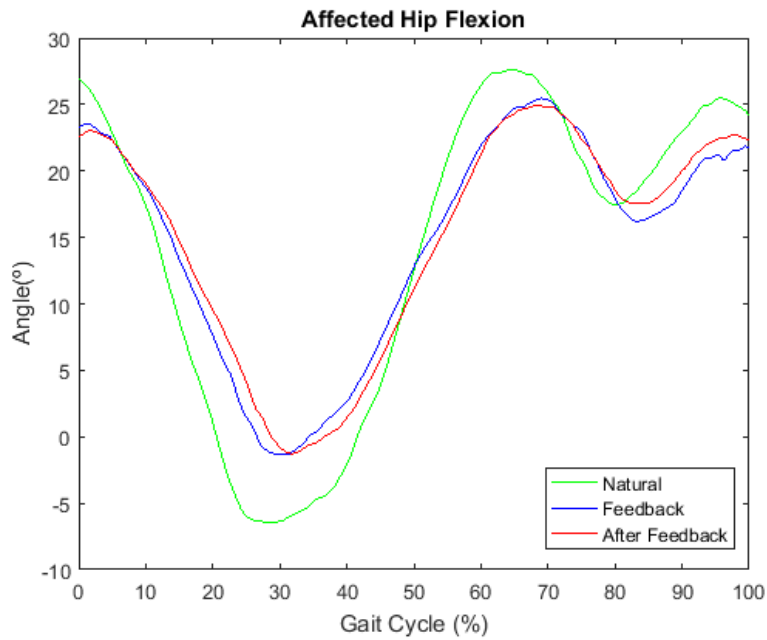


Fig. 36. Non-affected hip flexion angle in a GC

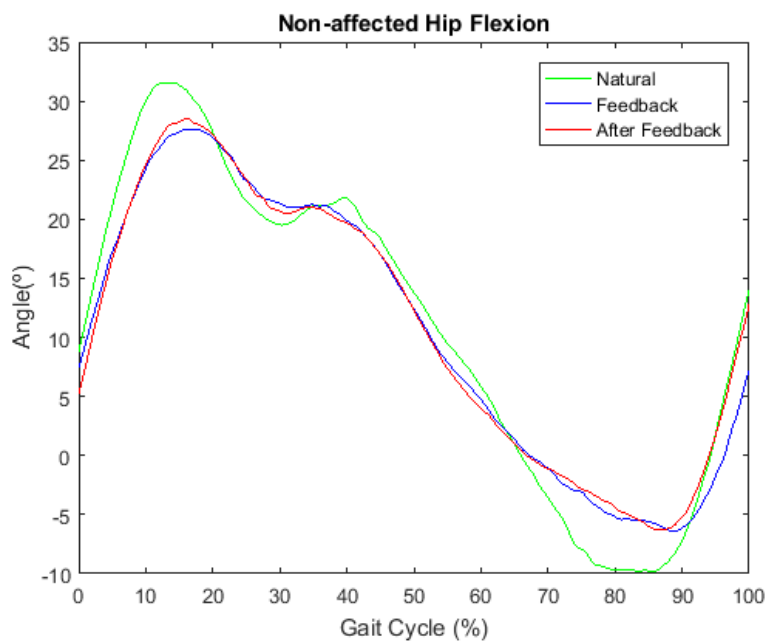


Fig. 37. Affected hip flexion angle in a GC

Coordinate	Angle type	Natural	Feedback	After feedback
Affected hip flexion	max	27,59	25,46	24,93
	min	-6,48	-1,37	-1,26
Non-affected hip flexion	max	31,60	27,65	28,58
	min	-9,84	-6,45	-6,27

Table 20. Maximum and minimum angles of hip flexion in a GC

Coordinate	Natural (N)	Feedback (F)	After feedback (A)	%F/N	%A/N
Affected hip flexion	34,07	26,83	26,20	-21,25	-23,10
Non-affected hip flexion	41,44	34,10	34,85	-17,71	-15,89

Table 21. ROM of the hip flexion angle

Coordinate	Natural-Feedback	Natural-After feedback	Feedback-After feedback
Affected hip flexion	3,97	4,34	1,16
Non-affected hip flexion	3,02	2,87	1,33

Table 22. RMSE between phases

In this case, the feedback was set on the left leg, because the patient was over flexing the non-affected hip (see Fig. 36). This is why the maximum angle for the left hip decreases with the use of the device. Concurrently, as seen in Table 20, the hip angle during stance reached $-6,5^\circ$, which showed an over extension, which is compensated with the feedback. Moreover, for both limbs, the ROM of the hip flexion decreases, which can be concluded as more control and stability of both limbs (Table 21).

When looking at the plots, an abnormality is seen in all the patterns. In the swing phase of the

GC of the affected leg, there is an unexpected double bump (seen from 60 to 95%), which is not solved with the use of the device. This is due to the pathology of the patient.

Satisfactorily, the change in pattern between feedback and after feedback is not significant, which is supported by the RMSE values (see Table 22). The gain of control in both limbs is visual, because the peaks seen in the natural gait decrease with feedback and after feedback.

8.3.2. Pelvic tilt

The pattern of the pelvic tilt angle in a GC is seen in Fig. 38:

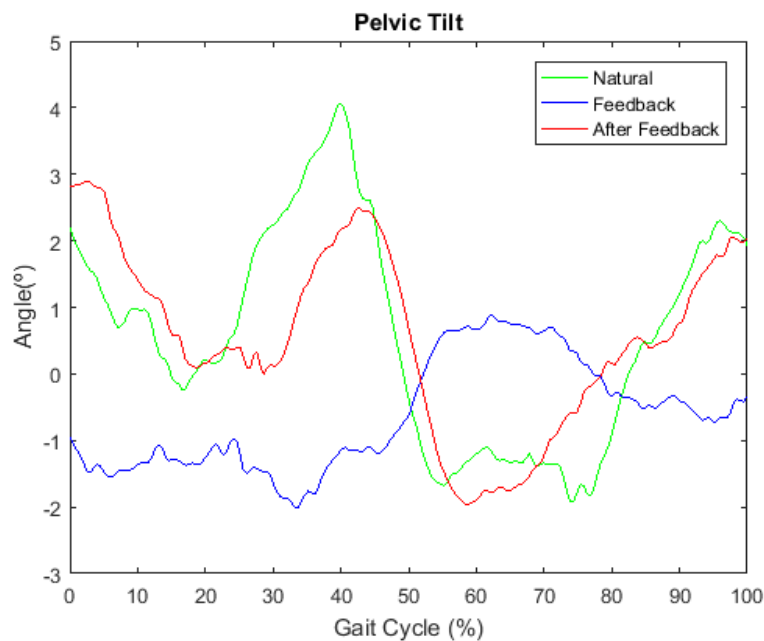


Fig. 38. Pelvic Tilt angle in a GC

Coordinate	Angle type	Natural	Feedback	After feedback
Pelvic tilt	max	4,06	0,89	2,90
	min	-1,91	-2,01	-1,97

Table 23. Maximum and minimum angles of pelvic tilt in a GC

Coordinate	Natural (N)	Feedback (F)	After feedback (A)	%F/N	%A/N
Pelvic tilt	5,97	2,90	4,87	-51,35	-18,32

Table 24. ROM of the pelvic tilt angle

Coordinate	Natural-Feedback	Natural-After feedback	Feedback-After feedback
Pelvic tilt	2,61	0,98	2,36

Table 25. RMSE between phases

There is a significant peak (maximum angle) during natural gait (Table 23), which is lowered with feedback. In addition, the ROM dramatically decreases with feedback (Table 24). It is also clearly seen that the pattern is closer to the standard one (see figure Fig. 38). This is directly related to gaining balance. When looking at the patterns and at the RMSE values (Table 25), the patient went back to the pelvic tilt angle pattern when stopping the feedback. Thus, it looks like he did not maintain the balance gained with feedback.

8.3.3. Knee flexion

Fig. 39 and Fig. 40 show the knee flexion angle pattern in a GC for both legs:

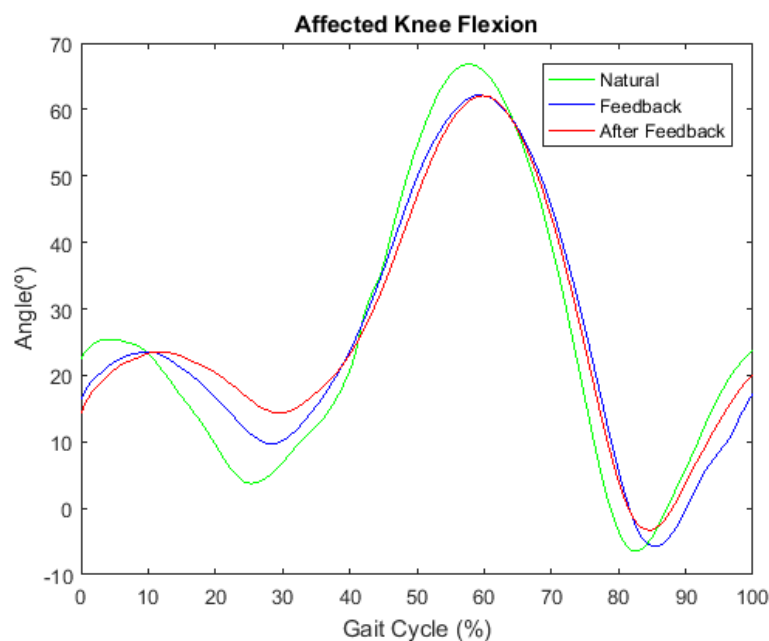


Fig. 39. Affected knee flexion angle in a GC

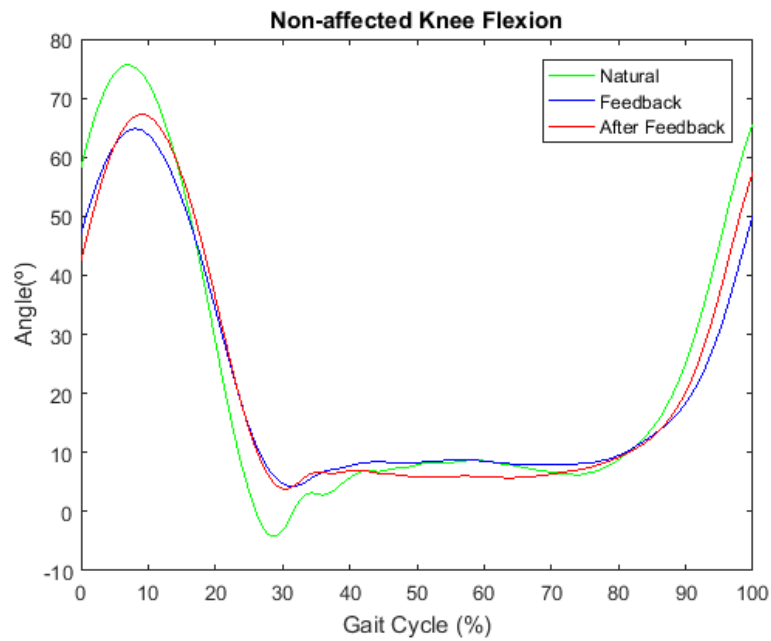


Fig. 40. Non-affected knee flexion angle in a GC

Coordinate	Angle type	Natural	Feedback	After feedback
Affected knee flexion	max	66,81	62,18	62,08
	min	-6,44	-5,80	-3,27
Left knee flexion	max	75,65	64,80	67,28
	min	-4,18	4,24	3,86

Table 26. Maximum and minimum angles of knee flexion in a GC

Coordinate	Natural (N)	Feedback (F)	After feedback (A)	%F/N	%A/N
Affected knee flexion	73,25	67,98	65,35	-7,20	-10,80
Non-affected knee flexion	79,83	60,56	63,42	-24,14	-20,56

Table 27. ROM of the knee flexion angle

Coordinate	Natural-Feedback	Natural-After feedback	Feedback-After feedback
Affected knee flexion	5,57	6,13	2,67
Non-affected knee flexion	6,90	5,94	2,73

Table 28. RMSE between phases

As seen in Table 26, both knees present hyperextension in natural gait. Whereas it is neutralized with feedback for the non-affected knee, it remains for the affected one.

As it happened with the hip flexion, the ROM decreases for both knees (Table 27). Whereas the ROM for the affected knee was wider than the one for the non-affected knee, they are more similar with the use of the device. Actually, the ROM of the non-affected knee is the one that changes more, which would be expected, since it was the limb under feedback. This shows how the patient gains control during the gait.

The non-affected knee presents a decreased ROM of the knee throughout the entire GC compared to the standard one. In the swing period, there is a diminished and delayed peak knee flexion (notice the flat pattern), which prevents not only normal limb clearance but also the foot from being placed in the proper position prior to the next foot contact [8]. This is called stiff-knee pattern. Even though the feedback was set on this limb, the stiffness remains with the use of the feedback.

When looking at Table 28, the changes in pattern with feedback are maintained in after feedback. Moreover, when looking at the pattern in both Fig. 39 and Fig. 40, the changes are visually seen. It shows more control of knee flexion.

8.3.4. Ankle dorsiflexion

In Fig. 41 and Fig. 42, the angle pattern for ankle dorsiflexion for both legs in a GC is seen:

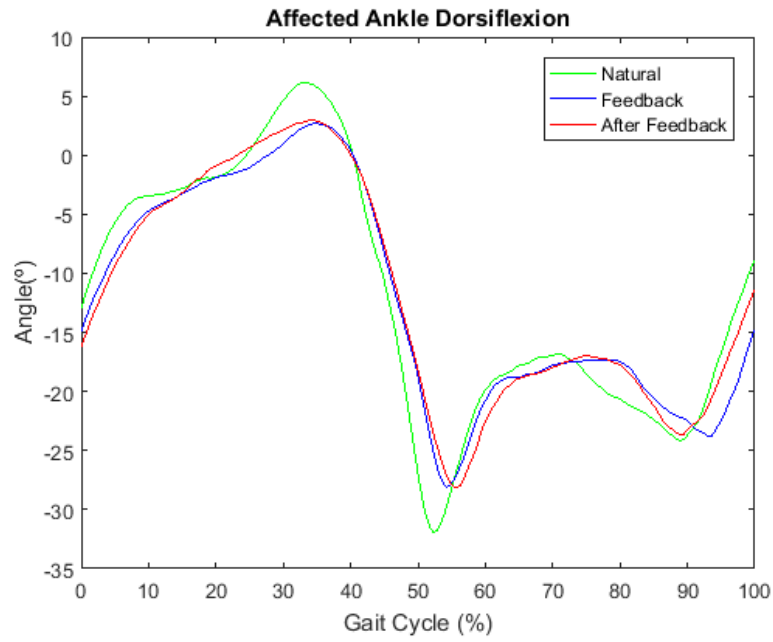


Fig. 41. Affected ankle dorsiflexion angle in a GC

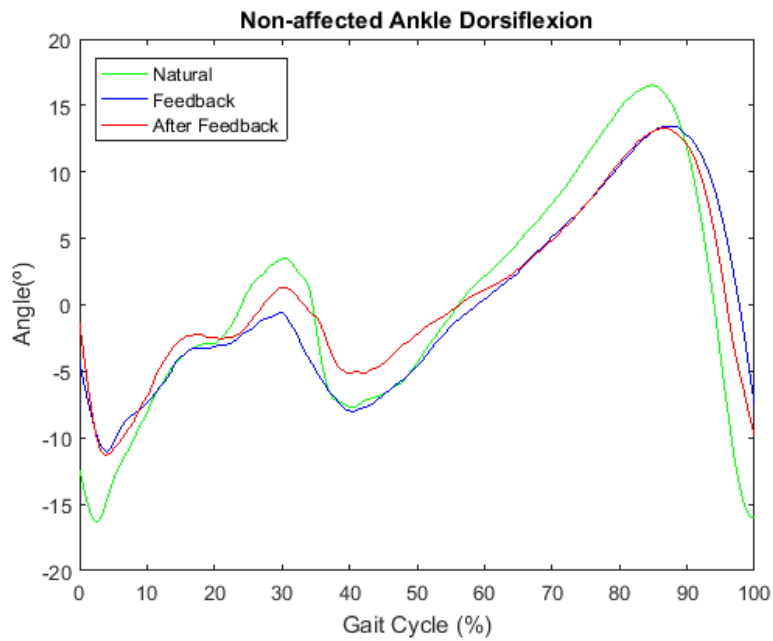


Fig. 42. Non-affected ankle dorsiflexion angle in a GC

Coordinate	Angle type	Natural	Feedback	After feedback
Affected ankle dorsiflexion	max	6,17	2,69	3,01
	min	-31,90	-28,06	-28,17
Non-affected ankle dorsiflexion	max	16,51	13,42	13,29
	min	-16,37	-11,06	-11,32

Table 29. Maximum and minimum angles of ankle dorsiflexion in a GC

Coordinate	Natural (N)	Feedback (F)	After feedback (A)	%F/N	%A/N
Affected ankle dorsiflexion	38,07	30,75	31,18	-19,23	-18,08
Non-affected ankle dorsiflexion	32,88	24,48	24,61	-25,56	-25,16

Table 30. ROM of the ankle dorsiflexion angle

Coordinate	Natural-Feedback	Natural-After feedback	Feedback-After feedback
Affected ankle dorsiflexion	2,96	2,83	1,35
Non-affected ankle dorsiflexion	4,12	3,34	1,89

Table 31. RMSE between phases

The values of the angles are not that different from one state to the other (Table 29). However, the ROM decreases in both cases, which is again connected to gaining control and stability (Table 30).

First of all, by having a look at natural gait pattern of the non-affected ankle, there is an unexpected peak on the swing phase (approx. 2% of the GC in Fig. 42). This peak decreases with feedback, but without disappearing. For both limbs, the ankle also gains control, but keeping an abnormal behaviour compared to the standard one.

As with the other coordinates, the pattern of the non-affected limb changes more than the affected one (Table 31). It may be interesting to remember that this patient tends to use a Foot-Up on his affected toe, but did not use it during the experimental trials.

8.4. Isabel

Patient Isabel was put under two feedbacks; one of 0° angle (F1) and one of 20° (F2) of the right thigh (considered as hip flexion).

8.4.1. Hip flexion

Fig. 43 and Fig. 44 show the hip flexion angle pattern for both legs in a GC:

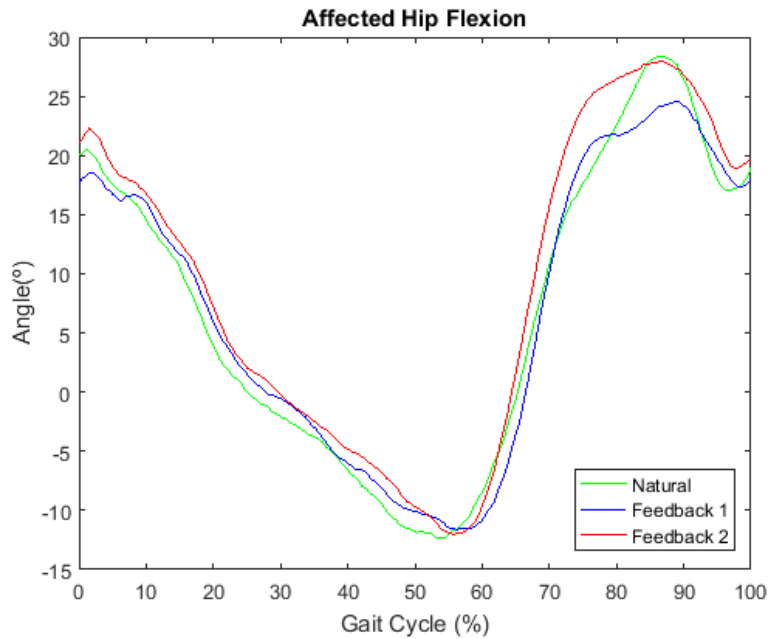


Fig. 43. Non-affected hip flexion angle in a GC

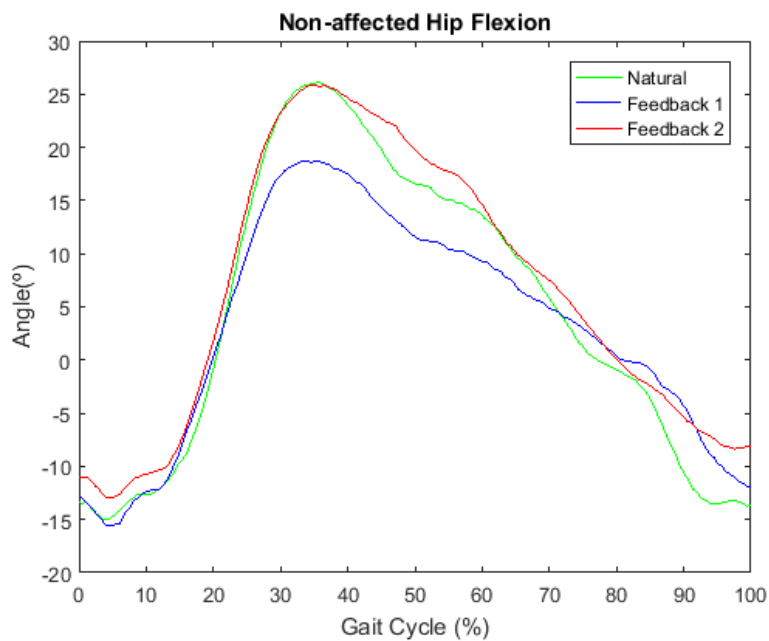


Fig. 44. Affected hip flexion angle in a GC

Coordinate	Angle type	Natural	Feedback 1	Feedback 2
Affected hip flexion	max	28,37	24,58	27,97
	min	-12,35	-11,63	-12,01
Non-affected hip flexion	max	26,11	18,76	25,85
	min	-15,02	-15,58	-13,01

Table 32. Maximum and minimum angles of hip flexion in a GC

Coordinate	Natural (N)	Feedback 1	Feedback 2	%F1/N	%F2/N
Affected hip flexion	40,72	36,22	39,98	-11,06	-1,83
Non-affected hip flexion	41,13	34,34	38,86	-16,51	-5,52

Table 33. ROM of the hip flexion angle

Coordinate	Natural-Feedback1	Natural-Feedback2
Affected hip flexion	1,85	2,70
Non-affected hip flexion	3,94	2,65

Table 34. RMSE between phases

There is an interesting issue with the ROM (see Table 33). For feedback 1, even though more for the non-affected limb, it decreases in both limbs and their final ranges of motion are similar. Even though less significantly, it also decreases for feedback 2, also more for the non-affected limb. For the peak values, there is not an important difference in feedback 2 (see Table 32). However, the maximum angle for both limbs decreases in feedback 1, which is the one with the angle set to 0°.

As commented above, the maximum angle decreases in feedback 1, but, when looking at the RMSE value, the difference is greater for the non-affected hip (Table 34). Given the condition of instability of the patient, feedback 1 seems to give her more control.

On the other hand, feedback 2 softens the peak in the terminal swing of the GC for the hip of the affected limb, which approaches the standard hip flexion pattern.

8.4.2. Pelvic tilt

The pattern of the pelvic tilt angle in a GC is seen in Fig. 45:

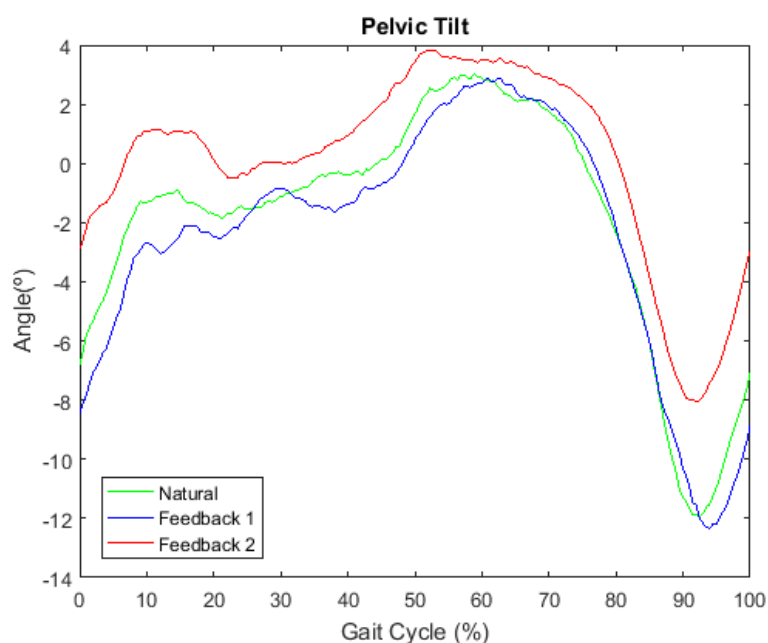


Fig. 45. Pelvic Tilt angle in a GC

Coordinate	Angle type	Natural	Feedback 1	Feedback 2
Pelvic tilt	max	3,03	2,91	3,82
	min	-11,92	-12,34	-8,05

Table 35. Maximum and minimum angles of pelvic tilt in a GC

Coordinate	Natural (N)	Feedback 1	Feedback 2	%F1/N	%F2/N
Pelvic tilt	14,95	15,24	11,87	1,97	-20,59

Table 36. ROM of the pelvic tilt angle

Coordinate	Natural-Feedback1	Natural-Feedback2
Pelvic tilt	0,97	2,22

Table 37. RMSE between phases

When looking at Fig. 46, the pattern of feedback 2 looks similar to the other ones, but shifted upwards. This is why the maximum angle of feedback 2 is greater (Table 35). This patient had grave stability problems, which can be seen in the pelvic tilt pattern, which is far from the standard one (shown in section 3.1). She could not walk for too long without resting.

For this reason, it is not surprising that the pelvic tilt pattern did not experience a great change with any of the feedbacks, as can be seen by looking at the RMSE values (Table 37). In However, when looking at Table 36, there is a great change in ROM with feedback 2. Thus, it could be a sign of gaining stability, which is exactly what the patient needs.

8.4.3. Knee flexion

Fig. 46 and Fig. 47 show the knee flexion angle pattern in a GC for both legs:

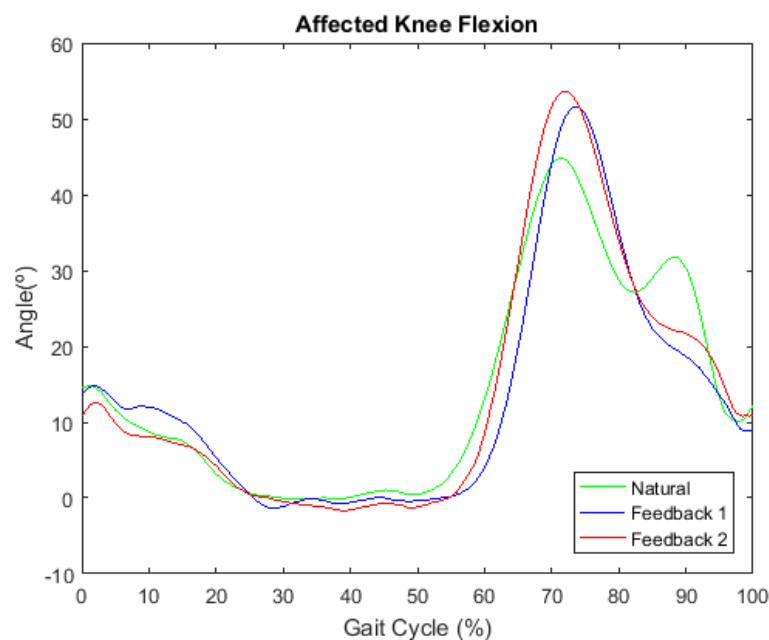


Fig. 46. Affected knee flexion angle in a GC

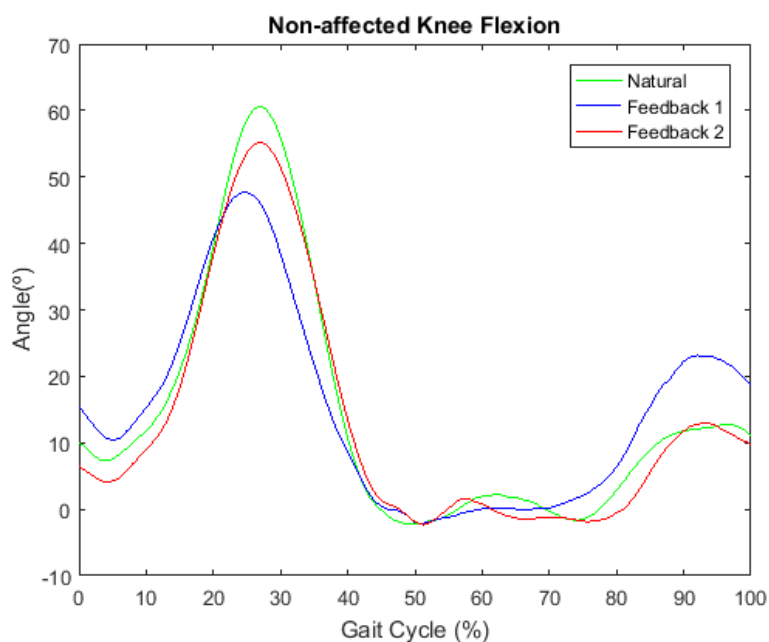


Fig. 47. Non-affected knee flexion angle in a GC

Coordinate	Angle type	Natural	Feedback 1	Feedback 2
Affected knee flexion	max	44,82	51,63	53,68
	min	-0,09	-1,43	-1,67
Non-affected knee flexion	max	60,58	47,71	55,24
	min	-2,30	-2,22	-2,32

Table 38. Maximum and minimum angles of knee flexion in a GC

Coordinate	Natural (N)	Feedback 1	Feedback 2	%F1/N	%F2/N
Affected knee flexion	44,91	53,06	55,35	18,13	23,23
Non-affected knee flexion	62,88	49,93	57,56	-20,59	-8,46

Table 39. ROM of the knee flexion angle

Coordinate	Natural-Feedback1	Natural-Feedback2
Affected knee flexion	5,24	4,05
Non-affected knee flexion	6,91	2,53

Table 40. RMSE between phases

It seems like there is compensation between both knees. Whereas the affected one gains ROM, the non-affected one decreases it, thus both ranges become closer in values (Table 39). Thus, gait becomes more symmetric. For feedback 2, the ROM are similar for both knees. Parallel, when looking at Table 38, it is seen that maximum angle of the affected knee increases, then flexing more, whereas the non-affected knee maximum angle decreases. In feedback 2, both knees flex at a similar maximum angle.

The most relevant difference, which is seen visually, is that the double peak present in the swing phase of the GC for the affected limb almost disappears with both feedbacks (see Fig. 46). For the affected limb, the knee flexion pattern is more similar to the standard one when using both feedbacks. For the non-affected knee, feedback 1 is the one that affects the most in terms of pattern, reducing more than 20% the knee flexion ROM. Also, the RMSE values (Table 40) show that both feedbacks seem to modify knee flexion pattern in a similar way for the affected limb. In the case of the non-affected limb, feedback 1 modifies pattern more significantly (also see Fig. 47).

8.4.4. Ankle dorsiflexion

In Fig. 48 and Fig. 49, the angle pattern for ankle dorsiflexion for both legs in a GC is seen:

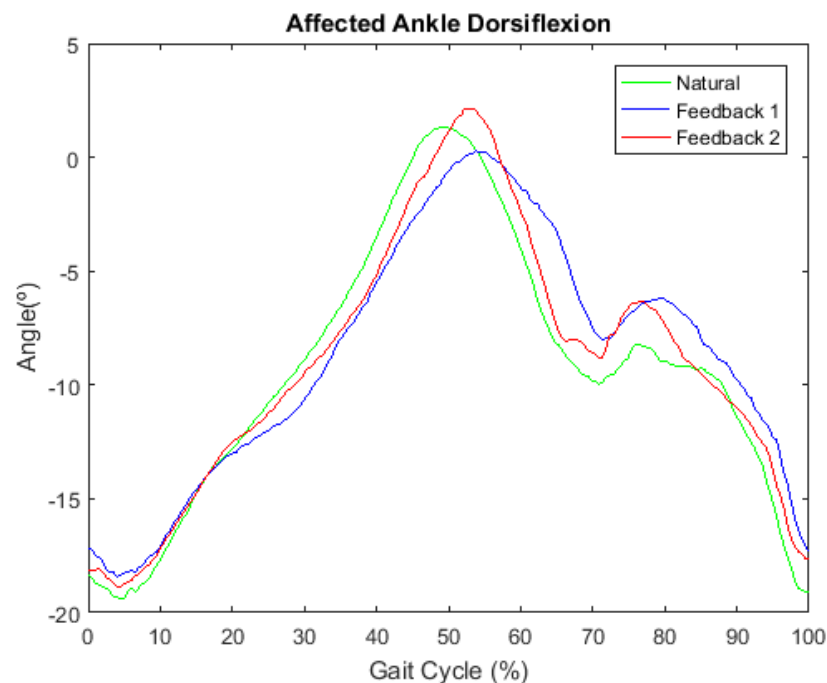


Fig. 48. Affected ankle dorsiflexion angle in a GC

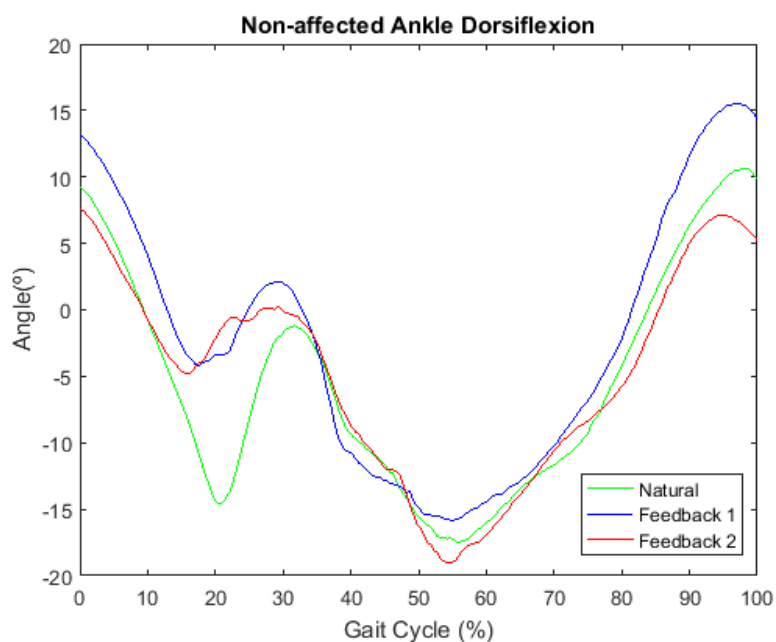


Fig. 49. Non-affected ankle dorsiflexion angle in a GC

Coordinate	Angle type	Natural	Feedback 1	Feedback 2
Affected ankle dorsiflexion	max	1,33	0,30	2,17
	min	-19,42	-18,43	-18,86
Non-affected ankle dorsiflexion	max	10,65	15,48	7,56
	min	-17,50	-15,83	-19,04

Table 41. Maximum and minimum angles of ankle dorsiflexion in a GC

Coordinate	Natural (N)	Feedback 1	Feedback 2	%F1/N	%F2/N
Affected ankle dorsiflexion	20,76	18,73	21,03	-9,78	1,32
Non-affected ankle dorsiflexion	28,14	31,32	26,60	11,29	-5,47

Table 42. ROM of the ankle dorsiflexion angle

Coordinate	Natural-Feedback1	Natural-Feedback2
Affected ankle dorsiflexion	2,06	1,15
Non-affected ankle dorsiflexion	4,28	3,53

Table 43. RMSE between phases

Table 42 shows that the ROM of the non-affected ankle increases with feedback 1 and slightly decreases with feedback 2, which does not modify it that much. Therefore, feedback 1 seems to provide less stability, since the peak values create a pattern further from the standard one.

The plot of the affected ankle shows a completely different pattern from the standard one. This is due to the fact that, as mentioned in section 5.3, the patient wears a DAFO, which significantly affects the kinematic behaviour of that joint.

The non-affected ankle presents a low peak on the initial swing (at approx. 20% of the GC; see Fig. 49), which decreases with both feedbacks. Both feedbacks seem to have the same effect on the non-affected limb during the GC, except from the transition from stance to swing of the GC, which seems to be better controlled for feedback 2.

As seen in Table 43, the RMSE values for the affected ankle show that both feedbacks have not modified the pattern, which is logical when thinking of the DAFO. On the other hand, the non-affected ankle dorsiflexion pattern is modified with both feedbacks, especially when it comes to the peak mentioned above.

9. Conclusions

The present Final Project for the Bachelor's Degree in Industrial Technology Engineering has analysed gait kinematics of paediatric patients with hemiparesis. It has been performed using the UPC Biomechanics Laboratory located at ETSEIB. Thus, there has been a learning process for the motion captures, along with all the used software: Motive, Matlab, and OpenSim. A musculoskeletal model has been adapted from one developed by the OpenSim team.

There has been a biomedical involvement in the project. The focus has been on paediatric hemiparesis, which is a form of cerebral palsy. Learning about the pathology and the effects that it has on the patients has been crucial to better understand their gait behaviour. The project has taken a different view than that of a regular engineering study, since these pathologies have different effects on each patients and each of them must be treated separately. The four angular coordinates that have been of interest for the study (pelvis tilt, hip flexion, knee flexion, ankle dorsiflexion) define orientations in the sagittal plane.

A prototype device, Walking o'Clock, has been analysed for the first time in the present project. Therefore, the aim of the work has been to analyse how the device modifies the gait pattern of each patient.

The multibody model has reproduced the motion of all the captures, as close to reality as possible. Some of them presented problems, but most of them could be solved by changing some parameters in the model. In case of not finding a solution, the capture was substituted with another one, because more than three were taken for every phase.

When having all the captures processed, gait cycles have been selected for each of them. After, the three chosen cycles for each phase have been merged to find a mean. Then, the gait cycles for each phase (natural, feedback and after feedback in case of patients Maria and Jordi; natural, feedback 1 and feedback 2 for patient Isabel) have been compared. The maximum and minimum angles have been calculated, as well as the ROM for the four studied coordinates. Also, the RMSE has been calculated to compare the three patterns.

The results of the analysis can be found satisfactory. The fact that the use of the device changes gait pattern has been confirmed. The feedback used for each patient was different and chosen by the physiotherapist. Thus, it modifies gait in different ways according to the patient. The objective was to approach standard or healthy gait during and after the use of the device.

The most interesting results of patient Maria are that she has gained ROM of the affected hip, while the non-affected has decreased. This can be a form of compensation. In after-feedback phase, she showed hypertension of the affected knee during stance phase, something that should be prevented, since it is a negative effect. Simultaneously, she presents a stiff knee gait in the natural phase, which is compensated with feedback. She also gains knee ROM for the

affected knee and the pattern approaches the standard one. For dorsiflexion of the affected ankle, which had an abnormal ROM, there is an important change, also approaching the standard pattern.

Patient Jordi has gained control while walking. The ROM has decreased in all four analysed coordinates, which can be concluded as more control over his limbs. The great decrease in the pelvic tilt ROM shows gain of stability. The hyperextension that he showed in the non-affected knee in natural gait disappears with feedback. It would be interesting to see how the device changes his gait while also using the Foot-Up.

Isabel has been studied differently, since she was set under two feedbacks, without checking the after feedback, due to her instability problems. What is interesting for knee flexion is that the double peak present in the swing phase of the GC for the affected limb almost disappears with both feedbacks and, the pattern is more similar to the standard one.

By looking at the results of the project, the idea was to predict how Walking o'Clock would work in future situations with other patients. After reporting the results to the Research Team of HSJD, they agreed on that this project can be a first step to proving that the use of the considered inertial device is positive. However, a deeper study should be done in the future. Bringing more patients to study would give a more general perspective with statistically significant results. Moreover, the timing should be greatly enlarged. Here, the feedback motion capture was taken after the patients had used it for a short time. It would be of more interest if this usage was of months, until the patient is completely adapted to it. The after effects should be then analysed after a stablished time, to see if the patient goes back to natural gait or maintains the one learned using feedback.

Which is interesting is that their pattern did not go back to the natural one when stopping the device. This means that, if used for a longer time, gait pattern could be modified permanently. In future situations, the physiotherapist, after being taught how to use the device and set feedback by the engineer, could use it in rehabilitation. They would have to find the most optimal feedback for the patient. Notice that Isabel used two different feedbacks, which had completely different results. Thus, it would take time and practise to find the best one for improving gait. Then, the device could be used by the patient in their daily basis, not only during rehabilitation at the clinic. It would be used as a gait assistant for as long as they needed. The physiotherapist would control their progression, and would modify the feedback progressively and according to the response of the patient. The simplicity of the device makes it reachable by any kind of patient.

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